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Studie zur neuromuskulären Stabilisierung des Sprunggelenkkomplexes anhand ausgewählter Muskeln

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- II. Wuebbenhorst, K. & Zschorlich, V. (2012). The effect of increasing external degrees of freedom on force production and neuromuscular stabilisation. J Spo Sci 30(14). 1561-1569.
- III. Wuebbenhorst, K. & Zschorlich, V. (2013). Interaction against different environmental dynamics during a leg extension task is controlled by temporal rather than amplitude scaling of muscular activity. J Electromyogr Kinesiol 23(5). 1029-1035.

Abstract

Das erfolgreiche Interagieren mit der Umwelt ist bezüglich der motorischen Kontrolle des Bewegungsapparates eine komplexe Aufgabe. Bei sich ändernden biomechanischen Anforderungen der Interaktion muss eine situationsadäquate Regulation der involvierten motorischen Strukturen erfolgen. Dazu müssen muskuläre sowie zentralnervöse Prozesse im Sinne der Aufgabenerfüllung aufeinander abgestimmt werden, um bspw. ein externes Objekt sicher kontrollieren zu können. Wird die Stabilität während der Aufgabenerfüllung nicht durch die Umwelt (bzw. durch das Interaktionsobjekt) gesichert, muss das neuromuskuläre System diese Aufgabe übernehmen. Dabei erfordert die gekoppelte Mechanik des Objektes mit dem motorischen Apparat spezifische muskuläre Beiträge zur Bewegung, welche sich je nach der Funktion des Muskels sowie der Aufgabe selbst in ihren Ausprägungen unterscheiden. Diese Arbeit ging der Frage nach, wie das neuromuskuläre System den Wechsel externer Bedingungen während einer Kraftproduktionsaufgabe kontrolliert. Dazu wurden an einem Bewegungsschlitten die möglichen mechanischen Freiheitsgrade (Degrees of Freedom; DoF) verändert und die Antworten des muskulären Systems mittels EMG von vier Muskeln (M. vastus medialis, M. tibialis anterior, M. peroneus longus, M. gastrocnemius medialis) registriert. Zusätzlich wurden dynamometrische Messungen vorgenommen und die resultierenden Kräfte in drei Dimensionen (F_x , F_y , F_z) erfasst und ausgewertet. Die EMG-Signale wurden mittels Kreuzkorrelationsberechnungen, Anstiegsuntersuchungen, Integral- sowie Differentialberechnungen und Spitzenamplitudenbestimmungen hinsichtlich intra- und intermuskulärer Koordinationsleistungen ausgewertet. Die dynamometrischen Daten wurden durch Integration, Differenzierung, Spitzenamplitudenbestimmungen sowie Kreuzkorrelationsuntersuchungen verarbeitet. Die Ergebnisse deuten auf eine Dominanz zeitlicher Einstellungsmechanismen der muskulären Aktivität gegenüber der Höhe der muskulären Aktivierung hin. Ferner zeigten sich intra- und intermuskuläre Koordinationsmuster als aufgabenspezifisch. Die Modulationen der Aktivitätsmuster der untersuchten Muskeln müssen in diesem Zusammenhang im Sinne einer Regulation der Gelenkstabilität (joint stiffness) gewertet werden. Da die Natur der Aufgabe eine multisegmentale Bewegung abverlangte, mussten die

einzelnen Subsysteme der kinetischen Kette individuell reguliert werden. So zeigte bspw. der M. vastus medialis einen Wechsel der Funktion von vorrangig kinetisch arbeitend hin zu einem Beitrag zur Stabilisierung des Knies, wenn die externen Freiheitsgrade größer und die Bewegung dadurch instabiler wurde. Allgemein war bei mehr DoF der Aufgabe (damit verbunden auch mehr Instabilität) ein Anstieg der muskulären Aktivitäten auszumachen. Dieser Mechanismus für die Bereitstellung der Gelenkstabilität wurde begleitet von einem Abfall der maximal exerzierbaren Kräfte (Integral als auch Spitzenwerte) und kürzeren elektromechanischen Verzögerungszeiten bei erhöhten DoF der Aufgabe. Diese Ergebnisse müssen im Sinne einer veränderten Koordination bei der Erzeugung stabilisierender als auch antreibender Kräfte innerhalb des motorischen Systems gewertet werden. Das Mehr an muskulärer Aktivität kann demnach nicht für die Krafterzeugung genutzt werden, sondern dient der Sicherung der Gelenkstabilität.

Zusammenfassend konnte belegt werden, dass Änderungen der mechanischen Umwelt zu spezifischen Adaptationen im Prozess der Gelenkstabilisierung und folglich der Sicherung dynamischer Stabilität führen. Die koordinativen Muster der untersuchten Muskulatur müssen als adaptiv an die jeweiligen Anforderungen der Aufgabe angesehen werden. Diese Adaptivität erfordert zum einen komplexe Regulationsvorgänge, ermöglicht jedoch auch eine Flexibilität in der Koordination multi-segmentaler Bewegungen. Schließlich muss hervorgehoben werden, dass zeitliche Koordinationsleistungen für die Sicherung eines möglich hohen Kraftoutputs als auch bei der Gewichtung muskulärer Beiträge eine entscheidende und dominierende Rolle einnehmen.

1. Einleitung und Zielstellung

Kennzeichnend für willkürliche Bewegungen des Menschen sind die hohe Variabilität und Adaptivität des motorischen Systems bei der Lösung motorischer Problemstellungen. Die Möglichkeit, sich auf verschiedene Umgebungskontexte einstellen zu können, die motorische Aktion entsprechend anzupassen und zielgerichtet der Erfüllung eines Ziels nachgehen zu können, stellt eine Grundlage für eine erfolgreiche Auseinandersetzung mit der Umwelt dar. Diese Grundfunktionen des motorischen Systems unterliegen vielfältigen Regelungs- und Steuerungsprozessen. Dazu findet die Wahrnehmung, Verarbeitung, Generierung und Weiterleitung bewegungsrelevanter Signale (Afferenzen und Efferenzen) in einem verteilten System der Motorik statt (Lemon, 1999). Die integrative Verarbeitung in allen Instanzen des Systems und die resultierende Ansteuerung der Effektoren (Muskeln), erlaubt in der Summe die Vielfältigkeit und Situationspezifität, welche für erfolgreiches Handeln notwendig ist.

Dem motorischen System inhärent sind komplexe Strukturen aus Knochen, Gelenken, Muskeln, Sehnen etc., welche eine Vielzahl an möglichen Freiheitsgraden (engl.: Degrees of Freedom; DoF) schaffen. Als einer der ersten erkannte Bernstein (1975) diese Zusammenhänge, weshalb die Koordination der großen Zahl an DoF des motorischen Systems auch als das Bernstein-Problem gilt (Arimoto & Sekimoto, 2011). Im Sinne der Aufgabenerfüllung stellen diese vielen DoF einen Vorteil in der Adaptivität an den jeweiligen Umweltkontext dar, schaffen aber zugleich durch die damit verbundene erhöhte Beweglichkeit (motor redundancy) auch Probleme in deren Kontrolle. Ferner sind im Bewegungsvollzug oftmals mehr DoF vorhanden, als für die Zielerfüllung gebraucht werden, welche in der Summe die Dimension des umgebenden Raumes übersteigen können (ebd.; Sporns & Edelman, 1993). Todorov & Jordan (2002) erklären diesbezüglich, dass diesem Konflikt zwei fundamentale Eigenschaften des motorischen Systems zugrunde liegen: zum einen, die Fähigkeit „high-level goals“ verlässlich und wiederholbar zu erreichen. Zum anderen, dass eine Variabilität bei den Bewegungs-Details existiert, welche bei einer erneuten Umsetzung einer Aufgabe verschiedene Freiheitsgrade (und somit unterschiedliche

Lösungsmöglichkeiten für ein gestelltes Problem, Anm. d. Verf.) involvieren kann (ebd.).

Bei der Auseinandersetzung mit der mechanischen Umwelt kommt der Kontrolle der Freiheitsgrade des motorischen Systems eine entscheidende Funktion zu. Im sportlichen Bewegungsvollzug können so mögliche Verletzungen verhindert werden, oder durch Stärkung kinetischer Ketten Höchstleistungen erzielt werden. Für das Erstgenannte wäre die notwendige Stabilität bei Absprung oder Landung während des Weitsprungs ein Beispiel. Ist diese nicht gewährleistet wie es z.B. nach Ermüdung der Fall ist, kann es zu Verletzungen kommen (Steib et al., 2013; Gribble & Hertel, 2004). Die adäquate Kontrolle der für eine Bewegung notwendigen Freiheitsgrade ist somit als elementarer Teil von Leistungen im sportlichen Handeln zu verstehen. Ergebnisse von Behm et al. (2002) zeigten diesbezüglich, dass bei Instabilität des Interaktionsmediums keine volle Aktivierung der kinetischen Muskulatur mehr möglich war und somit auch die maximal mögliche Leistung sank. Die Autoren argumentieren, dass zwei Stressoren des zentralen Nervensystems (Versuch der maximalen Krafterzeugung und Aufrechterhaltung der Stabilität) in ihrer Kombination die inhibitorischen Effekte bedingten. Folglich kann vermutet werden, dass der suffizienten Stärkung der Subsysteme kinetischer Ketten eine entscheidende Bedeutung zukommt.

Die gezielte Integration dieses Stresses in Interventionen oder Trainingsprogrammen muss als essenziell angesehen werden, um den Körper zu Adaptationsleistungen an neue Stimuli zu forcieren (Anderson & Behm, 2004). Diese Effekte werden bei der Implementierung sensomotorischer oder propriozeptiver Konzepte in Therapien bei der Rehabilitation nach Verletzungen (Gollhofer, 2003; Bruhn et al., 2004) oder beim athletischen Training ausgenutzt. Dabei sei in diesem Zusammenhang auf die komplexen Zusammenhänge mit sensomotorischen Regelkreisen und deren Bedeutung für die Ansteuerung der involvierten Muskulatur verwiesen (z.B. Woollacott et al., 1986; Goodworth & Peterka, 2012). Die Afferenzen von Muskelspindeln und anderen Rezeptoren sind als integrative Bestandteile muskulärer Aktivierungsmuster aufzufassen.

Die vielen Freiheitsgrade des motorischen Systems bedingen, dass das Nervensystem aus einem großen (unendlichen) Repertoire möglicher Lösungen wählen muss, da in der Regel nicht genau spezifiziert werden kann, wie und in

welcher Form jeder Muskel und jedes Gelenk kontrolliert wird (Ting 2007). In theoretischen Auseinandersetzungen wurden die Minimierung kinetischer „cost-functions“ (Mutha et al., 2008), optimale Feedback-Kontrolle (Todorov & Jordan, 2002) oder internale Modelle auf der Basis gepaarter „forward models“ und „inverse models“ (Kawato, 1999) vorgeschlagen¹. Im Zuge der geschilderten Überlegungen ergibt sich die Notwendigkeit, die Kontrolle der Freiheitsgrade, welche durch unterschiedliche Aufgabenstellungen oder mechanische Umweltkontexte veränderbare Anforderungen schafft, auf der Ebene des neuromuskulären Systems zu erbringen (Holl & Zschorlich, 2011). Dies gilt insbesondere für Aufgaben, welche sich nicht durch Stabilität auszeichnen bzw. bei der die Bewegungsstabilität nicht durch die Umweltinteraktion sichergestellt werden kann (z.B. Laufen über Waldboden vs. Laufen auf ebenen Boden). Den involvierten Gelenken kommt in diesem Zusammenhang die Aufgabe zu, die geplante Bewegung zu ermöglichen, aber auch nicht benötigte Freiheitsgrade (auch die anderer Gelenke) einzuschränken (Kornecki & Zschorlich, 1994). Die mit diesen Kontrollmodalitäten einhergehenden Anpassungsleistungen der Gelenk-Stabilität an die Anforderungen der Aufgabe stellen eine Voraussetzung für eine Robustheit der motorischen Leistungs-Variabilität und adäquaten Gelenksicherung gegen externe Perturbationen dar (Franklin et al., 2003).

Im sportlichen aber auch im alltäglichen Handeln ist ein unmittelbares Ziel, möglichst effektiv bzw. möglichst viel Kraft aufzubauen. Die physiologisch zugrunde liegenden Mechanismen involvieren zum einen Anpassungserscheinungen der Muskulatur wie Querschnittsvergrößerungen oder auch des Fiederungswinkels (Aagard et al., 2001) aber auch neuromuskuläre Adaptationen in efferenten als auch afferenten Bahnen (Holl & Zschorlich, 2011; Gollhofer, 2003). Solche funktionalen Adaptationen in den einzelnen Bereichen schaffen veränderte Voraussetzungen für die Kontrolle der Muskelaktivität und führen zu angepassten Leistungen („motor outputs“) im jeweiligen Handlungskontext. Bei der Bewertung des Beitrags eines Muskels zu einer Bewegung klassifizierte Nemessuri (1968) die Funktionen von Muskeln als Fixator (Gelenkstabilisierung), Kinetor (Antriebsfunktion) oder Modulator (Bewegungsregulator). Zschorlich (1987) ergänzt, dass diese Differenzierung nicht

¹ Auf eine nähere Auseinandersetzung mit Modellen motorischer Kontrolle wird an dieser Stelle verzichtet.

immer eindeutig abgrenzbar ist und auch die Funktion während des Bewegungsvollzugs wechseln kann.

Wie bereits mehrfach erwähnt, besitzt der jeweilige Umweltkontext entscheidende Einflüsse auf die Anforderungen einer Aufgabe und somit auf die Umsetzung dieser. Das Resultat sind aufgabenspezifische neuromuskuläre Kontrollmechanismen, welche sich in den verschiedenen Ebenen des motorischen Kontrollsystems manifestieren.

Diese Synopsis verdeutlicht die Effekte veränderter mechanischer Umweltbedingungen auf neuromuskuläre Anpassungsleistungen anhand einer ballistischen, unilateralen Beinextensionsbewegung. Im Fokus stehen kontextspezifische neuromuskuläre Kontrollmodalitäten als grundlegendes Regelungssystem bei der Lösung verschiedener Aufgaben.

2. Methodische Hintergründe und Fragestellungen

2.1. Apparatur und Aufgabe

Methodisch basieren die dieser Arbeit zugrunde liegenden Versuche auf einem Wechsel der mechanischen Bedingungen eines externen Objekts („Bewegungsschlitten“) mit welchem die Probanden interagierten (s. Abb. 1). Der Bewegungsschlitten ermöglichte durch eine veränderte Fußauflage die Schaffung dreier mechanischer Zustände, welche jeweils durch die sukzessive Einführung von einer größeren Zahl an DoF unterschiedliche Anforderungen an die Stabilisierung bedingten. Die Krafterzeugung gegen den Schlitten durch die Probanden war somit durch unterschiedlich stark kompromittierte Stabilitätszustände der gekoppelten Systeme (motorisches System des Probanden mit externem mechanischem System) gekennzeichnet. Folglich schufen die variierend instabilen Interaktionsbedingungen unterschiedliche Grade der notwendigen Stabilisierungsleistung durch das motorische System. Infolgedessen sind Veränderungen auf neuromuskulärer Ebene als auch bei der Leistung (motor output) zu erwarten.

Zu diesem Zweck wurde zwischen drei Bedingungen in diesen Versuchen unterschieden². Eine mechanisch stabile Bedingung wurde kreiert, indem der Bewegungsschlitten rigide gemacht und eine feste Fußauflage für die Probanden installiert wurde. Diese Bedingung erlaubte keine Änderung in den Gelenkwinkeln und keine Nutzung der Freiheitsgrade von Knie- und Sprunggelenk. Diese Bedingung galt als statisch und wurde als ‚0 DoF‘-Bedingung bezeichnet. Für eine zweite Bedingung wurde der Schlitten in der z-Ebene (vgl. Abb. 1) beweglich gemacht, die rigide Fußplatte jedoch beibehalten. Diese Ebene erlaubte eine aktive Bein-Extensionsbewegung unter stabilen Fußauflagebedingungen und wurde als ‚1 DoF‘-Bedingung definiert. In einer dritten Bedingung wurde die rigide Fußplatte durch eine instabile ersetzt. Diese wurde über ein Kugelgelenk mit dem Schlitten verbunden und erlaubte Bewegungen mit 3 DoF. Diese 3 Freiheitsgrade beinhalteten horizontale, rotationale und translatorische Bewegungsmöglichkeiten, welche Inversion/Eversion und Dorsal-/Plantarflexion des Sprunggelenks erlaubten.

² Für eine genaue Beschreibung des genutzten Bewegungsschlittens sei auf die Originalliteratur verwiesen. In dieser Synopsis wird auf eine detaillierte Darstellung verzichtet.

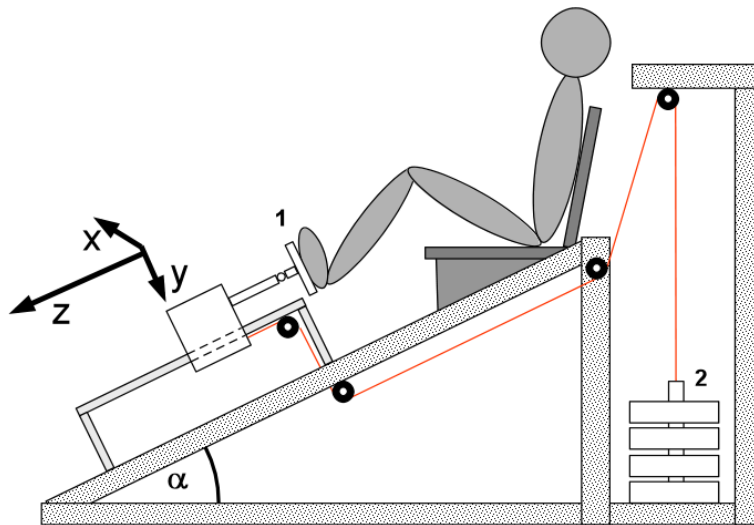


Abbildung 1 Schema des Bewegungsschlittens und den definierten Kraft- und Bewegungsdimensionen. 1- variierende Fußauflage, 2- Gegengewicht

Der Schlitten stellt in seiner Funktionsweise eine geschlossene kinetische Kette (gkK) dar. Die Nutzung einer gkK ist aus biomechanischer und neuromuskulärer Sicht mit Anforderungen während athletischer Bewegungen vergleichbar (Escamilla et al., 1998). Ferner wurde durch die hohe Intensität (maximale ballistische Extensionsbewegungen) jedes einzelnen Versuchs bezweckt, den Anstrengungen während natürlich vorkommender Bewegungen nahe zu kommen (Willardson, 2004). In dieser Studie bot sich die Möglichkeit, standardisierte Ausgangsbedingungen zu schaffen. Dies wurde durch die Kontrolle der initialen Knie- und Sprunggelenkwinkel mittels eines manuellen Goniometers umgesetzt (ca. 90° jeweils). Der Proband befand sich während der Versuche auf dem Schlitten in einer sitzenden, an seine Größe angepassten Position. Der Widerstand des Gegengewichts (Position 2 in Abb. 1) betrug 37.5 kg.

Durch dieses Setup war es möglich, die Stabilisierungsmechanismen in Isolation von posturalen Kontrollmechanismen zu studieren. Mögliche konfundierende Effekte durch die Sicherung posturaler Stabilität wurden so weitgehend eliminiert. Folglich sind die Einstellungsleistungen auf neuromuskulärer Ebene als Antwort auf die

verschiedenen DoF-Bedingungen und nicht durch bewegungsinduzierte sensorische Diskrimination anzunehmen.

Zusätzlich konnte durch die Nutzung der drei verschiedenen Freiheitsgradbedingungen zwischen bewegungsinduzierten und stabilisierungsbedingten Anpassungen des motorischen Systems unterschieden werden.

Aus diesen Methodischen Vorüberlegungen heraus wurden zwei grundlegende Fragestellungen abgeleitet:

- 1) Welchen Effekt besitzt eine variierende Anzahl an Freiheitsgraden bei der Kraftproduktionsaufgabe auf dem Bewegungsschlitten?
- 2) Wie sehen die neuromuskulären Anpassungsleistungen bei der Stabilisierung der Aufgabe unter den jeweiligen mechanischen Zuständen des externen Systems aus?

2.2. Messung der Effekte auf das neuromuskuläre System und der Kräfte

Zur Abschätzung der Auswirkungen der unterschiedlichen biomechanischen Anforderungen der Freiheitsgradbedingungen auf die Muskulatur wurden elektromyographische Messungen an vier ausgesuchten Muskeln der rechten unteren Extremität vorgenommen. Dazu wurden der M. peroneus longus (PL), M. tibialis anterior (TA) als Vertreter der sprunggelenksumspannenden Muskeln, der M. gastrocnemius medialis (GM) als biartikulärer Muskel mit Ursprung am Knie und Insertion in die Achillessehne sowie der M. vastus medialis (VM) als Teil des extensorischen M. quadriceps femoris ausgewählt. Zu näheren Einzelheiten der Messmethodik wird auf die Originalliteratur verwiesen.

Die EMG-Signale wurden mittels der Kreuzkorrelation (Wuebbenhorst & Zschorlich, 2011) und EMG-Anstiegsuntersuchungen (Wuebbenhorst & Zschorlich, 2013) für die Bestimmung zeitlicher Muster als auch mittels Integration (Wuebbenhorst & Zschorlich, 2012) sowie Amplitudenberechnungen zur Messung der Aktivität weiterverarbeitet (Wuebbenhorst & Zschorlich, 2013). Durch die Kombination dieser Analyseverfahren war es möglich, zwischen zeitlichen und aktivitätsbestimmten Strategien, sowie intra- und intermuskulären Koordinationsleistungen des

motorischen Systems bei Lösung der Bewegungsaufgaben zu differenzieren. Für die Kreuzkorrelation wurden dazu die Laufzeitdifferenzen in Bezug zu einem Referenzmuskel (M. vastus medialis) bestimmt.

Die Beurteilung der Kraftparameter erfolgte mittels der Dynamometrie. Dazu wurde ein KISTLER-Kraftaufnehmer verwendet, welcher sich direkt im Kraftweg (Fz-Richtung, s. Abb.1) befand. Der Kraftaufnehmer erlaubte die Messung dreier Kraftdimensionen (Fx, Fy, Fz), welche entsprechend der Abb. 1 definiert wurden (Fz entspricht somit der Kraft in Bewegungsrichtung; direktionale Kraftkomponente). Anhand der gemessenen Kraftdaten wurden die Parameter Maximalkraft, Impuls (Integral der Kraft-Zeit-Kurve), Kraftänderungsgeschwindigkeit (maximales Differential der Kraft-Zeit-Kurve) sowie in Kombination mit der Kreuzkorrelationsanalyse der EMG-Daten auch die elektromechanische Verzögerungszeit berechnet.

2.3. Studiendesign

Die Studie wurde ausschließlich mit sportlich aktiven Männern im Alter von 21 bis 33 Jahren durchgeführt. Die Probanden erhielten vor den Versuchen eine schriftliche sowie eine mündliche Aufklärung über die Versuche und deren Hintergründe. Keiner der Probanden hatte jemals zuvor Versuche auf dem Bewegungsschlitten durchgeführt. Ferner war zum Testzeitpunkt jeder Proband gesund und keiner wies irgendeine Form chronischer Sprunggelenksdefizite, neurologische Auffälligkeiten, Achillessehnenprobleme oder sonstige Beeinträchtigungen auf.

Durch die homogene Struktur der Probandenpopulation wurden bewusst Einschnitte der externen Validität der Studie toleriert. Ferner besteht die Möglichkeit einer „healthy worker bias“. Folglich können die Effekte von Alter, Geschlecht oder physischer Status durch die aktuellen Experimente nicht effektiv beurteilt werden. Die Resultate anderer Studien welche u.a. das Geschlecht berücksichtigen (Norwood et al., 2007) befinden sich jedoch in einer Linie mit diesen Experimenten. Das Design erlaubte jedoch, dass jeder Proband im Vergleich der Freiheitsgrade sein eigener Kontrollproband war. Somit waren die inner-Subjekt-Faktoren (within-subject factors: DoF-Bedingung, Kraftkomponente, Muskelparameter) für alle Probanden gleich. Der Einbezug von Zwischensubjektfaktoren (between-subject factors) entfiel in diesen

Experimenten. Durch dieses Design konnte eine Reduktion der interindividuellen Variabilität durch die Nutzung nur einer Gruppe erreicht werden.

3. Forschungsstand und Einbettung der eigenen Arbeiten

Die Interaktion mit der mechanischen Umwelt besitzt entscheidende Einflüsse auf die Regelungs- und Steuerungsprozesse des motorischen Systems. In der Theorie internaler Modelle³ (Wolpert & Kawato, 1998; Kawato, 1999) finden die physikalischen Eigenschaften des umgebenden Raumes als auch die des Interaktionsobjektes als kontextuales Signal Eingang in die motorische Antwortselektion. Dieses Signal liefert Informationen über die äußeren Zustände (Objekt-Physik & Umgebungs-Physik) unter denen Bewegung produziert werden soll. Im Sinne der Theorie der internalen Modelle kann, basierend auf dem Wissen um die äußeren mechanischen Verhältnisse, durch die Spezifizierung von vorwärts gerichteten („forward-model“) und inversen Modellen eine gezielte Umsetzung der Zielbewegung erfolgen. Im Falle des Greifens eines Objektes kann so das Verhältnis von „grip force“ zu „load force“ so gestaltet werden, dass sie dem Objekt selbst angemessen ist (ebd.). Dies setzt jedoch auch die Kenntnis der physikalischen Eigenschaften des Objektes (Masse, Größe, Form, Oberflächenbeschaffenheit, Dehnbarkeit etc.) voraus. Ist die Bewegung selbst durch externe Kräfte beeinflusst (Umgebungs-Physik), z.B. durch Wirkung eines externen, perturbierenden Kraftfeldes, muss das neuromuskuläre System für die Kompensation sorgen. Im theoretischen Sinne der internalen Modelle kann die Regulation der Bewegung durch ein Update spezifischer Module erfolgen. So kann gelernt werden, durch veränderte Kontrollmodalitäten des motorischen Systems auf veränderte Voraussetzungen der Umweltinteraktion sowie der Objekt-Physik zu reagieren.

³ In dieser Arbeit wird nicht der Frage der Gültigkeit der verschiedenen Modelltheorien nachgegangen.

Den theoretischen Grundlagen der Modelle liegen komplexe neuromuskuläre sowie zentralnervale Veränderungen zugrunde. Diese Veränderungen sind grundlegend an die informationsverarbeitenden Systeme des sensomotorischen Systems gekoppelt. Die situationsadäquate Verarbeitung der sensorischen Informationen und die Implementierung dieser in die Bewegungsausführung sind als eine wesentliche Voraussetzung für erfolgreiches motorisches Handeln anzusehen.

Bei der Kontrolle verschiedener externer Freiheitsgrade während einer Interaktion mit einem Objekt (oder Interaktionsmedium) kommen diesen Einstellungsmechanismen entscheidende Bedeutungen zu. Bober et al. (1982) untersuchten neuromuskuläre Adaptationen bei einer Kraftproduktionsaufgabe der oberen Extremitäten gegen einen Handgriff mit unterschiedlicher Beweglichkeit (durch Variation der Freiheitsgrade des externen Objekts). Die Autoren unterschieden zwischen Bedingungen bei maximaler Kraftproduktion und 40% sowie 60% der Maximalkraft. Sie zeigten, dass bei instabilen Interaktionsbedingungen im Vergleich zu stabilen mehr Aktivität der stabilisierenden Muskeln des Handgelenks für die Aufgabe notwendig waren. Dieser Effekt war jedoch geringer ausgeprägt bei 40% und 60% der Maximalkraft. Ferner konnte nachgewiesen werden, dass die Sequenz der Muskelaktivierung durch die stabilisierenden Muskeln des Handgelenks initiiert wurde. Das Mehr an Kraft, sowie die frühere Rekrutierung distaler gelenkstabilisierender Muskulatur dient der notwendigen Stabilisierung der biokinematischen Kette des Arms bei stärkerer Kraftproduktion durch die antreibende Muskulatur. Die aktive Stabilisierung ermöglichte die Einschränkung der funktionellen Freiheitsgrade des Handgelenks, um möglichst effektiv Kraft aufbauen zu können.

Kornecki et al. (1992) ergänzt, dass die Komposition des motorischen Systems selbst (bestehend aus Gelenken, Muskeln, Bändern etc.) eine Vielzahl von Freiheitsgraden schafft (mindestens 30 an den oberen Extremitäten, ebd.). Die resultierenden Bewegungsmöglichkeiten (motor redundancy) des motorischen Systems bewirken, dass neben den gewünschten auch unerwünschte Bewegungen entstehen. Zur zielgerichteten Umsetzung einer Aufgabe wird jedoch zu jeder Zeit nur ein bestimmter Prozentsatz der potentiell verfügbaren Mobilität benötigt. Demnach kommt der Einschränkung der ungewollten DoF durch Stabilisierungsmechanismen eine entscheidende Funktion zu (ebd.). Die Implementierung biologischer, dem motorischen System inhärenter Beschränkungen („constraints“, Gielen et al. 1995)

erlaubt nach Kornecki (1992) die Blockade einiger Gelenke bei gleichzeitiger gekoppelter Bewegung in anderen Gelenken.

Diese ‚constraint induced redundancy reduction‘ (van Soest & van Galen, 1995) erlaubt die Koordination des Zusammenspiels verschiedener involvierter Gelenke, wirft jedoch auch die Frage nach den zugrunde liegenden Mechanismen der Regulation auf. Gielen et al. (1995) gibt als eine Möglichkeit die Implementierung der constraints auf neuronaler Ebene an. Dies beinhaltet die aufgabenspezifische Aktivierung der Muskeln entsprechend ihres Beitrags zur Bewegung (Kinetor, Stabilisator, Modulator). Durch die Spezifität der Muskelaktivierung können beteiligte Subsysteme einer kinetischen Kette selektiv kontrolliert und so die notwendige Reduktion der involvierten Freiheitsgrade erwirkt werden (van Soest & van Galen, 1995; Wübbenhorst & Zschorlich, 2013). Dieses Prinzip ist dadurch gekennzeichnet, dass die constraints selbst-implementiert sind (van Soest & van Galen, 1995). Im Gegensatz dazu können Beschränkungen der Freiheitsgrade auch durch die Wirkung physikalischer Prinzipien erwirkt werden (z.B. Rotations-Translations-Kopplung) (ebd.). Letzteres sei der Vollständigkeit halber nur erwähnt, wegen der geringen Relevanz für die Arbeit jedoch nicht näher vertieft. Die physikalischen implementierten „constraints“ machen das System weniger beweglich, während die selbst-implementierten Beschränkungen nur als Teillösung angesehen werden können, welche die vielen Freiheitsgrade überwinden sollen, diese jedoch nicht reduzieren (ebd.). Kornecki (1992) benennt daher die Stabilisierung selbst als einen Prozess, welcher durch aktive muskuläre Beschränkungen zeitlich redundante DoF innerhalb des motorischen Systems kontrolliert.

Bei der Interaktion mit der Umwelt sind jedoch nicht nur die Freiheitsgrade des motorischen Systems entscheidend. Zusätzlich müssen auch die physikalischen Bedingungen des Interaktionsmediums in eine Bewegung einbezogen werden. Bei mechanisch instabilen Interaktionen addieren sich zu den DoF des motorischen Systems die des Mediums. Daraus resultieren besondere Anforderungen an die Regelung der Mechanik der involvierten Gliedmaßen. Bei mehrgelenkigen Bewegungen muss eine aufgabenspezifische Regulation der Gelenkstiffness durch das Nervensystem erfolgen, um die möglichen Freiheitsgrade des Bewegungsapparates zu kontrollieren. Diese Kontrolle muss jedoch unabhängig von den für die intendierte Bewegung benötigten Kräfte funktionieren (Shemmel et al.,

2010; Hogan, 1985). Auf neuronaler Ebene kann eine solche Kontrolle durch feedforward-Signale erfolgen, wodurch aufgaben- und anforderungsspezifische Muster der Kokontraktion involvierter Muskeln resultieren (Franklin et al., 2004; Holl & Zschorlich, 2011). Diese Kokontraktionsmuster bieten die Möglichkeit, schnell externen Perturbationen durch die Aufgabe entgegen zu wirken, bedingen durch die anhaltenden Kontraktionen jedoch auch, dass bestimmte metabolische Kosten damit einhergehen (Franklin et al., 2004). Die Koordination von sensorischem und motorischem System erlaubt, dass auf der Grundlage der gekoppelten Mechanik von Umgebung und Akteur (bzw. dessen Gliedmaße) eine möglichst große Kraftproduktion erfolgen kann (Kornecki & Zschorlich, 1994). Verändern sich durch Bewegung die biomechanischen Voraussetzungen, so kann durch feedforward und feedback-Prozesse eine adaptive Regulation der aktuellen Bewegung und damit der stiffness-Kontrolle erfolgen (Krutky et al., 2010). Durch eine derartige spezifische Kontrolle der Muskulatur, der Gelenke und deren Zusammenspiel kann eine Bewegung stabilisiert werden, wenn die Umgebung die Stabilität entweder kompromittiert bzw. die Stabilität nicht selbst ermöglicht (Perreault et al., 2008).

Die Mechanismen, welche die Grundlage solcher Koordinationsleistungen bilden, sind jedoch weitgehend ungeklärt. Viele Arbeiten untersuchten willkürliche Stabilisierungsaufgaben an den oberen Extremitäten. Die vorliegende Arbeit zielt vorrangig auf eine nähere Klärung der Mechanismen bei willkürlichen Stabilisierungsaufgaben der unteren Extremitäten ab. Aus diesem Grund wurden Versuche mit der unter Punkt 2 beschriebenen Methodik durchgeführt. Die dieser Synopsis zugrunde liegenden Originalarbeiten zielen auf verschiedene Aspekte der neuromuskulären Stabilisierung des Sprunggelenkes ab. Folgende Publikationen dienen als Fundament:

Wuebbenhorst K., Zschorlich V. (2011): Effects of muscular activation patterns on the ankle joint stabilization: an investigation under different Degrees of Freedom. *J Electromyogr Kinesiol*, 21(2). 340–347.

Wuebbenhorst, K. & Zschorlich, V. (2012): The effect of increasing external degrees of freedom on force production and neuromuscular stabilisation. *J Spo Sci*, 30(14). 1561-1569.

Wuebbenhorst, K. & Zschorlich, V. (2013): Interaction against different environmental dynamics during a leg extension task is controlled by temporal rather than amplitude scaling of muscular activity. *J Electromyogr Kinesiol*, 23(5). 1029-1035.

In den folgenden Darstellungen werden die wichtigsten Ergebnisse dieser Studien in den aktuellen Forschungsstand eingebettet. Für einzelne Nachweise wird auf die Originalarbeiten verwiesen. Die folgende Tabelle gibt einen Überblick über die analysierten Parameter in den einzelnen Publikationen.

Tabelle 1 Übersicht der dieser Synopsis zugrunde liegenden Artikel

Jahr	Journal	Getestete DoF-Bedingungen	Analyseparameter
2011	Journal of electromyography & kinesiology (5 year impact: 2.178)	0,1,3	Kreuzkorrelationsanalyse von EMG und Kraft; Muskelaktivitätsmessung mittels Integral, Maximalkraftmessungen
2012	Journal of Sports Sciences (impact factor: 2.1)	0,1,3	Maximalkraftmessungen, Kraftanstiegsrate, Muskelaktivitätsmessung via Integral, Berechnung prozentualer Unterschiede
2013	Journal of electromyography & kinesiology (5 year impact: 2.178)	1,3	EMG-Anstiegsuntersuchungen, Muskelaktivitätsmessungen mittels Amplitude und Integral, Kraftmessung mittels Integral

Ein wichtiger Teil allgemeinen motorischen Handelns ist es, entsprechend der Aufgabe ausreichend Kraft zu produzieren. Bei sportlichen Bewegungen bestimmt das entsprechende Kraftausmaß oftmals über den Sieg oder die Niederlage. Ferner

kann durch ein adäquates Maß an Muskelaktivierung möglichen Verletzungen bei biomechanisch veränderten Umgebungsbedingungen (z.B. Wechsel des Untergrundes beim Laufen) vorgebeugt werden. Bei inadäquatem Verhältnis von bewegungs-(bzw. kraft-) und stabilisierungsregulierendem System ist eine suffiziente Sicherung der Gelenkintegrität oftmals erschwert (Bruhn, 2003). Die notwendigen Synchronisierungsleistungen beider Systeme werden zentral erbracht und beinhalten unter anderem Modelle sensorischer Dynamiken, körperlichen Voraussetzungen, dem Handlungsplan sowie externer Bedingungen (Frank & Earl, 1990). Der motorische Output (Muskelaktivierung, Krafterzeugung) ist als Summe solcher Einstellungsleistungen aufzufassen.

Wie bereits dargestellt, muss das motorische System bei der Realisierung von Bewegungen zwei getrennte Leistungen vollbringen: 1) Sicherstellung der für die Freiheitsgradreduktion benötigten Muskelaktivierung und 2) Erzeugung der für die intendierte Bewegung benötigten Kräfte. Diese Kombination muss so koordiniert werden, dass die Stabilität aller involvierten Gelenke nicht durch eine überschießende Aktivierung der Muskulatur kompromittiert wird (Holl & Zschorlich, 2011). Auf neuronaler Ebene kann durch die Regulation der Wirkung bewegungsinduzierten peripheren Feedbacks z.B. durch verstärkte präsynaptische Inhibition einer solchen überschießenden Aktivierung entgegengewirkt werden. Auch eine verstärkte Transmission in kortikospinalen Leitungsbahnen kann entscheidende Einflüsse dabei besitzen (ebd.). Bei mechanischer Instabilität zeigten sich diese Effekte deutlich und waren besonders zu Beginn der Bewegung ausgeprägt (ebd.).

Der Effekt von Instabilität auf die Kraftproduktion und –übertragung auf die Umwelt zeigte sich bei Untersuchungen an den oberen Extremitäten als verringert (Bober et al., 1982; Kornecki, 1992; Kornecki & Zschorlich, 1994; Kornecki et al., 2001). Die Effektivität der Kraftübertragung des Akteur-Objekt-Systems wurde mit 80% beziffert, bei zusätzlicher Stabilisierung des Objektes (Kornecki & Zschorlich, 1994). Die Funktion (antreibend oder stabilisierend) der involvierten Muskeln sowie deren Zusammenspiel beeinflusst die Wirkung auf die einzelnen Subsysteme der kinetischen Kette. Werden bestimmte Synergien verändert, kann der Einfluss eines Muskels an einem Subsystem größer oder kleiner werden (Chvatal et al., 2011). Entsprechend muss das Zusammenspiel mehrerer Muskeln bei veränderten Bedingungen reorganisiert werden. Folglich ist eine effektive Stabilisierung möglich,

wenn die an einem Gelenk erzeugten Drehmomente von zueinander antagonistischen Muskeln balanciert sind (Kornecki & Zschorlich, 1994). Unter Bedingungen von Instabilität zeigten sich aufgrund der veränderten mechanischen Bedingungen veränderte, an die Aufgabe adaptierte Koordinationsmuster der Muskulatur (Wuebbenhorst & Zschorlich, 2011). Bezüglich der Kontrolle der Kraftproduktion auf externe Objekte vermuteten Kornecki et al. (2001), dass die Gleichgewichtslage und nicht die Mobilität des Objektes entscheidend sind bei der Determination der Kräfte, welche durch einzelne Muskeln erzeugt werden.

Die neuromuskuläre Aktivierung muss dementsprechend so abgestimmt sein, dass die Koordinationsmuster der Bewegung den variierenden biomechanischen Bedingungen adäquat sind. Dies kann u.U. durch Regulation von Rekrutierungsmustern, elektromechanischen Verzögerungszeiten und Muskelaktivitäten bewerkstelligt werden (Wuebbenhorst & Zschorlich, 2011). Diese Rekonfiguration der Muskelaktivitäten (Hogan, 1985) zeigte sich spezifisch für die jeweilige Aufgabe (0, 1 oder 3 DoF) (Wuebbenhorst & Zschorlich, 2011). Dabei erwies sich in Analogie zu den oberen Extremitäten, dass die stabilisierende Muskulatur des Sprunggelenks bei mehr Instabilität (3 DoF-Bedingung) eine frühere Rekrutierung aufwies (ebd.). Im Falle des M. peroneus longus war zudem eine engere Kopplung mit dem M. vastus medialis (VM) bei der Stabilisierung von 3 DoF auszumachen. Dies kann im Sinne einer synergistischen Kopplung zur Stärkung der Subsysteme „Knie“ und „Sprunggelenk“ als Teile der gesamten kinetischen Kette der abverlangten Versuchsbewegung gewertet werden. Im Falle des VM (als Teil der Quadrizeps-Muskulatur eigentlich mit antreibender Funktion) steht zu vermuten, dass ein Beitrag zur Kniestabilisierung bei Situationen mit großer Instabilität erfolgte, welche bei mehr Stabilität der mechanischen Interaktion selbst nicht vorhanden war. Der Anstieg der Muskelaktivität des VM trotz geringerer effektiver Kraftübertragung auf den Bewegungsschlitten bekräftigt diese Vermutung (ebd.). Ferner äußerte sich die steigende Stabilisierungsarbeit bei Erhöhung der DoF des Schlittens in geringeren elektromechanischen Verzögerungszeiten (electromechanical delay, EMD). Dies konnte für alle drei Krafft Dimensionen (F_x , F_y , F_z) bestätigt werden. Es wird vermutet, dass das neuromuskuläre System dadurch günstige Ausgangsbedingungen für eine schnelle Kraftproduktion (als Voraussetzung für die Stabilisierung) schafft (ebd.).

Es muss ergänzt werden, dass die Änderungen der Aktivitätsprofile der Muskeln mit weiteren Adaptationsmechanismen einhergehen. Bei der Sicherung der Stabilität von Gliedmaßen spielen neben den willkürlichen Muskelaktivierungsmechanismen auch unwillkürliche Antworten eine Rolle (Hasan, 2005). Die Modulation Ia-afferenten Feedbacks als Folge antagonistischer Kokontraktion sprunggelenksstabilisierender Muskeln zu Beginn und während der Bewegungsausführung auf dem Schlitten gilt ebenfalls als modulierender Einfluss (Holl & Zschorlich, 2011). Zusammenfassend lässt sich konstatieren, dass die Gelenkstabilisation als Erscheinungsform der Bewegungskoordination gewertet werden muss (Bruhn, 2003).

Die veränderten zeitlichen Rekrutierungsmuster zusammen mit den adaptierten Muskelaktivitäten (gemessen über das integrierte EMG, iEMG) waren auf intra- als auch intermuskulärer Basis nachweisbar (Wuebbenhorst & Zschorlich, 2011, 2013). Dies dokumentiert die komplexen Vorgänge bei der Ansteuerung der Muskulatur unter dem Einfluss variierender Stabilisierungsbedingungen. Die Analyse der iEMGs liefert jedoch lediglich Informationen über die Gesamtaktivität eines Muskels über einen definierten Zeitraum, erlaubt jedoch keine Aussagen über die Charakteristik der entsprechenden Muskelaktivität. Zu diesem Zweck wurden die Antworten der untersuchten Muskeln auf Stabilisierungsanforderungen anhand der Charakteristika der Aktivitätsprofile ausgewertet (Wuebbenhorst & Zschorlich, 2013).

Analog zur Analyse der Rekrutierungsprofile (Wuebbenhorst & Zschorlich, 2011, s. oben) unterlagen auch die Aktivitätscharakteristika situationsspezifischen Veränderungen (Wuebbenhorst & Zschorlich, 2013). Während die iEMGs bei 3 DoF im Vergleich zu 1 DoF für VM und TA signifikant erhöht waren, zeigte die Spitzenaktivität (peak electrical activity, PEA) des EMGs keine Veränderungen für die untersuchten Muskeln. Somit konnten keine Adaptationen der Amplitude des EMGs eines Muskels (PEA), aber eine Veränderung der Aktivität über die gesamte Zeit (iEMG) der Bewegung nachgewiesen werden. Ferner erwies sich jedoch die Zeit bis zum Erreichen der PEA (time to peak, TTP) als variabel für 3 der 4 Muskeln (mit Ausnahme des TA). Für GM, PL und VM waren jeweils längere TTP Zeiten nachweisbar. Die Analyse des EMG-Anstiegs als Parameter des Gesamtsystem-Verhaltens zeigte sich folglich muskelspezifisch verändert im Vergleich von 1 & 3 DoF (ebd.). Diese Ergebnisse zeigen, dass eine Stiffness-Regulation auf der Basis einer erhöhten Ko-Kontraktion entsprechender Muskeln bewerkstelligt wird (siehe

iEMG-Ergebnisse). Den Amplitudenänderungen kann jedoch durch eine abgestimmte zeitliche Koordination (PEA: keine Änderung, TTP: starke Änderungen) entgegengewirkt werden. Eine genaue Dosierung der Kraft eines Muskels zu bestimmten Zeitpunkten zeigte sich somit als dominante Strategie beim Stabilisierungsprozess. Die Nutzung einer zeitlichen Gradierung gegenüber einer Spitzenamplitudenveränderung zeigte sich als favorisiert in dieser Studie und stützt somit die Impuls-timing-hypothese von Corcos et al. (1993) (ebd.).

Die Kontrolle der intra- als auch intermuskulären Rekrutierungsmuster (Wuebbenhorst & Zschorlich, 2011) sowie der Charakteristika der Aktivitätsprofile (Wuebbenhorst & Zschorlich, 2013) erhöht somit die durch das motorische System zu kontrollierenden Variablen im Stabilisierungsprozess, erlaubt jedoch ein gewisses Maß an Flexibilität in der Koordination mehrgelenkiger Bewegungen (Gomi & Osu, 1998). Diese Flexibilität ließ sich vorrangig durch Variation zeitlicher Muster nachweisen. Der Regulation von insbesondere der am Sprunggelenk ansetzenden Muskeln (TA, PL) kommt eine besondere Bedeutung zu, da diese Muskeln die Stabilität über dem Subtalar-Gelenk sichern sollen, indem exzessive Gelenkrotationen verhindert werden (Neptune et al., 1999).

Die aufgezeigten Einstellungsmaßnahmen als Adaptationen des motorischen Systems an die externen Bedingungen dienen der Aufhebung des destabilisierenden Effektes der Kraftproduktion selbst (Rancourt & Hogan, 2009). Veränderte Stiffness-Eigenschaften durch Aktivitätsmodulationen der involvierten Muskeln schaffen aber auch andere Voraussetzungen für die Kraftproduktion. Bober et al. (1982) vermuteten, dass eine Beeinträchtigung der Kraftproduktion durch den Stabilisierungsprozess auf der Grundlage einer verringerten Effektivität von afferenten und efferenten Ansteuerungsvorgängen als auch des Timings zustande kommt. Deutliche Veränderungen konnten durch die bisher zitierten Studien in den Ansteuerungsprozessen (Holl & Zschorlich, 2011) als auch im Timing (Wuebbenhorst & Zschorlich, 2011, 2013) belegt werden.

Wie bereits erwähnt sank die Effektivität der Kraftproduktion auf ca. 80% bei Stabilisierungsanforderungen während Bewegungen der oberen Extremitäten (Kornecki & Zschorlich, 1994). Die Untersuchungen der unteren Extremitäten am Bewegungsschlitten zeigten ähnliche Werte der Effektivität von 76.3% (Abfall um 23.7%) bei Betrachtung der Kräfte in Bewegungsrichtung (Fz-Dimension) im

Vergleich von 0 und 3 DoF (Wuebbenhorst & Zschorlich, 2012). Die Untersuchung der Querkräfte in vertikaler (Fy-Dimension) und horizontaler (Fx-Dimension)(s. Abb. 1 im Methodenteil) Richtung zeigten Abfälle der effektiven Kraftproduktion von bis zu 41.6% (Effektivität = 58.4%) bei Fy und 33.5% (Effektivität= 66.5%) bei Fx, wenn stabilisiert werden musste. Zusätzlich dokumentierte die Analyse des Kraftanstiegs („rate of force development“), dass bei mehr DoF der Aufgabe ein signifikanter Abfall zu verzeichnen war (ebd.). Die Ergebnisse belegen, dass ergänzend zu den neuromuskulären Adaptionsleistungen des Stabilisierungsprozesses auch die Parameter der Kraftproduktion beeinflusst sind. Die geringeren Spitzenkräfte bei mehr Instabilität sind somit als Antwort auf die zu leistende Stabilisierungsarbeit zu sehen. Diese Sichtweise ist in Analogie zu den Betrachtungen der oberen Extremitäten (Bober et al. 1982).

Der Abfall, der durch die Stabilisierungsanforderungen beeinflussten Maximalkraftwerte, steht in unmittelbarem Zusammenhang mit der Kraftanstiegsrate. Aagaard et al. (2002) hoben in diesem Zusammenhang hervor, dass die Kraftanstiegsrate einer der Hauptdeterminanten der Maximalkraft während schneller, ballistischer Bewegungen ist. Die gefundene Reduktion des Kraftanstiegs bei mehr DoF der Aufgabe deutet in diesem Fall auf eine Strategie des motorischen Systems hin, welche auf einer verringerten Interferenz mit dem Prozess der Krafterzeugung beruht (Wuebbenhorst & Zschorlich, 2012). Gestützt wird diese Annahme dadurch, dass eine Erhöhung der Freiheitsgrade auch einen progredienten Abfall der Kraftanstiegsrate bedingte (ebd.). Zusätzliche Unterstützung findet dieser Ansatz dadurch, dass bei Betrachtung des Integrals der Kraftkurve („force impulse“) ein Abfall von 1 auf 3 DoF von lediglich 6.8% gefunden wurde (Wuebbenhorst & Zschorlich, 2013) wohingegen die Analyse der Maximalkräfte einen Abfall von 20.3% zeigte (Wuebbenhorst & Zschorlich, 2012). Folglich, kann vermutet werden, dass das neuromuskuläre System die Verluste für die Kraftproduktion über die Zeit der Bewegung so effektiv wie möglich versucht zu kompensieren. Eine solche Strategie erlaubt die (möglichst effektive) Aufrechterhaltung der Fähigkeit Kraft auch unter instabilen Bedingungen gegen ein Interaktionsobjekt auszuüben (Wuebbenhorst & Zschorlich, 2013).

In allen durchgeführten Studien zeigten sich Reduktionen der Maximalkräfte in allen drei Dimensionen (Fx, Fy, Fz; s. Abb. 1), wenn die externen Bedingungen instabil

wurden. Diese Kraftverluste sind in den Kontext der Gelenkstabilisierung einzuordnen. Wird eine Kraft in eine Richtung aufgebracht, wird eine reaktive Kraft in entgegengesetzter Richtung, jedoch mit gleichem Betrag produziert (Seo & Armstrong, 2009). Die reaktiven Kräfte wiederum erzeugen externe Momente am jeweiligen Gelenk, welche durch interne Momente des gleichen Betrags im Sinne eines Gleichgewichts ausgeglichen werden müssen (ebd.). Dazu nutzt das motorische System die an einem Gelenk ansetzende Muskulatur mit antagonistischer Wirkung und moduliert, wie in den Untersuchungen dargestellt, deren Aktivität entsprechend des jeweiligen Kontextes. In stabilen Situationen können die externalen Momente reduziert werden und die Stabilisierungsanforderungen sind geringer (Seo & Armstrong, 2009).

Für die Erzeugung eines möglich großen Betrages an Kraft gibt es jedoch einen mechanischen Vorteil durch die Kopplung der Kraft in Bewegungsrichtung (F_z) und der Querkräfte (F_y , F_x) (ebd.). Können Kräfte auch in andere Richtungen als nur in die Bewegungsrichtung (direktionale Kraft, F_z) selbst erzeugt werden, so kann eine Hebelwirkung („leverage“) genutzt werden, um auch die direktionale Kraft zu verstärken (ebd.). Da in den instabilen Situationen dieser Hebel-Effekt nicht so stark genutzt werden kann, sinken auch die F_z -Kräfte. Diese Zusammenhänge konnten empirisch in mehreren Studien belegt werden (z.B. Bober et al., 1982; Kornecki & Zschorlich, 1994; Seo & Armstrong, 2009; Wuebbenhorst & Zschorlich, 2011, 2012, 2013). Seo & Armstrong (2009) begründen den Effekt auf das motorische System damit, dass in den stabilen Situationen externalen Momente am Gelenk reduziert werden können und somit die benötigten Aufwendungen für eine adäquate Gelenkstabilisierung minimiert werden konnten. Dieser Effekt fehlt in den Situationen mit instabiler Interaktionsmechanik. Die dieser Synopsis zugrunde liegenden Studien erweitern die Überlegungen zu den grundlegenden Mechanismen dieses Effektes. So kann durch die Studien vermutet werden, dass der Grad der mechanischen Instabilität (0, 1 oder 3 DoF) den leverage-Effekt graduell abstuft. Diese Idee wird durch den progredienten Abfall aller drei Spitzenkraftwerte (F_x , F_y , F_z) bei Erhöhung der DoF gestützt (Wuebbenhorst & Zschorlich, 2013).

4. Zusammenfassung, Ausblick und Methodenkritik

Die dieser Synopsis zugrunde liegenden Studien konnten einen tieferen Einblick in die Mechanismen der Gelenkstabilität geben. Es konnten, in Analogie zu Untersuchungen der oberen Extremitäten, Zusammenhänge zwischen bewegungs- als auch stabilisierungsrelevanten muskulären Kontrollmechanismen bei der Umsetzung der Aufgabe dargestellt werden. Neben kontextabhängigen Variationen in den exerzierbaren Kräften und den Aktivitäten stabilisierender und antreibender Muskulatur sind insbesondere die Timing-Aspekte muskulärer Kontrolle als bestimmende Einflussgröße hervorzuheben. Der zeitlich koordinative Aspekt der Stabilisierung erwies sich gar als dominant gegenüber aktivitätsregulierenden Strategien (Wuebbenhorst & Zschorlich, 2013). Insgesamt stellte sich der Prozess der Gelenkstabilisierung auf dem Bewegungsschlitten als komplex und in seiner Regulation als hierarchisch dar. Letzterer Punkt beruht auf der Annahme, dass die Stärkung der biokinematischen Kette des Gesamt-Effektorsystems als notwendige Voraussetzung für eine erfolgreiche Interaktion mit den Umweltbedingungen angesehen werden muss. Diese Notwendigkeit bedingt Aufwendungen des motorischen Systems, welche durch die aufgabenbestimmte variable Implementierung von ‚non-holonomic-constraints‘ erwirkt werden können. Die verringerte Kräfteerzeugung auf die Umwelt ist dann als Ergebnis der Einstellungen des motorischen Systems in Abhängigkeit der extern wirkenden Bedingungen aufzufassen.

Die Regulation der einzelnen Subsysteme (der Gelenke) der genutzten biokinematischen Kette wurde in dieser Arbeit nicht explizit untersucht. Durch die beschriebenen Veränderungen der untersuchten Muskeln lassen sich jedoch komplexe, inhärent variable Einstellungsstrategien vermuten. Dieser Ansatz steht dabei im bedingten Gegensatz zu Studien von Nashner (1977), welcher bei der Stabilisierung des Standes feste Muster bei funktionell verbundenen Muskeln der unteren Extremität bei schnellen posturalen Reaktionen fand. In den dieser Synopsis zugrunde liegenden Studien konnte gezeigt werden, dass in Abhängigkeit von der Aufgabe bestimmte muskuläre Muster dominieren (z.B. frühzeitige Aktivierung stabilisierender Muskeln), diese Strategien im Gesamtkontext jedoch nicht als fest anzusehen sind. Die beschriebenen Ergebnisse ergänzen die gefundenen spezifischen Veränderungen in der Ia-afferenten und kortikospinalen Ansteuerung

sprunggelenksumfassender stabilisierender Muskeln (Holl & Zschorlich, 2011). Gleichzeitig erhöhen sich jedoch auch die notwendigen Kontrollvariablen zur Lösung der Aufgabe unter der jeweils herrschenden externen Mechanik.

Angesichts einer geringen externen Validität der Studie, durch die bisherige strenge Reglementierung der Inklusionskriterien für die Probanden, sind weitere Studien notwendig, um die Generalisierbarkeit der Ergebnisse zu untersuchen. Erste, bis dato jedoch unveröffentlichte Ergebnisse zeigen, dass das Alter als modulierende Größe bei der Kontrolle der Muskelaktivitätsmuster während der Stabilisierung einwirkt.

Auch die Natur der Aufgabe ist als elementar in der Ausprägung des Antwortverhaltens der Muskulatur anzusehen. Als Grundlage dienten unilaterale, ballistische Beinextensionen mit nur geringem Trainingshintergrund. Das Design erlaubte so jedoch, Ermüdungseffekte weitgehend auszuschließen. Ferner waren die Probanden stets über die jeweilige Versuchsbedingung informiert, sodass antizipatorische Mechanismen einen entscheidenden Einfluss gehabt haben können (Holl & Zschorlich, 2011). Gerade auf der Ebene spinaler Kontrolle kann dieses Vorwissen Einfluss auf die Ausprägung der Kontrollprozesse besitzen (ebd.). Weitere Studien könnten in diesem Zusammenhang klären, inwieweit die Vorinformation Einfluss auf Kontrollmuster der Muskulatur bei den einzelnen mechanischen Bedingungen besitzt.

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Effects of muscular activation patterns on the ankle joint stabilization: An investigation under different Degrees of Freedom

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ABSTRACT

Altered biomechanical demands of a stabilization task lead to specific changes in coordination patterns among the involved muscles. The objective of this work was to investigate the effects of increased Degrees of Freedom (DoF) of an external object on the stabilization process of the ankle joint in a voluntary force production task. Four muscles (vastus medialis, VM; tibialis anterior, TA; peroneus longus, PL; gastrocnemius medialis, GM) were recorded using surface electromyography and synchronized to dynamometric data. The subject's task was to exert force against the external object by performing a knee extension under 0, 1 or 3 DoF. Forces were measured using three dimensional force transducers and temporal coordination was assessed using the cross-correlation function (CCF). While the force decreased with increasing DoF the muscles showed a selective gain scaling in order to stabilize the ankle joint. Muscles fulfilling mainly stabilizing functions (TA and PL) tended to increase their activities, while the muscles with motor functions either decreased (GM) or increased (VM). The CCF revealed different intermuscular coordination strategies depending on the environmental condition, showing an advanced phasing in the ankle stabilizing PL in unstable environmental conditions (3 DoF). Nevertheless, the overall sequence of muscle activation was preserved. It is concluded that the process of joint stabilization is controlled in dependency of the status of the external system. The associated neuromuscular system adjustments underline the role of movement coordination in the stabilization process.

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1. Introduction

Everyday motor tasks are challenging to the motor system regarding the control of motor redundancy known as the Bernstein problem (Turvey 1990; Sporns and Edelman 1993). The composition of the motor apparatus of multiple linkages and joints and an even greater number of muscles can attain a large number of Degrees of Freedom (DoF), which can exceed the dimension of the workspace in which the organ operates (Sporns and Edelman 1993). The redundancy causes the nervous system to choose from a large set of possible solutions since the task demands are not sufficient to individually specify how each muscle and joint must be controlled (Ting 2007). In order to solve the process, it is suggested that the nervous system includes an optimization process, which aims on the minimization of kinetic cost functions (Mutha et al., 2008). However, when a different amount of DoF of an external object must be controlled, the neuro-muscular system is exposed to substantial changes in order to control the altered task demands. More specifically, when a force is applied to an external object with

differing equilibrium states the maximum of applicable force is decreased (Bober et al., 1982). Additionally, the loss of force is accompanied by changes in the contribution of different muscles to the movement.

Nemessuri (1968) has distinguished muscles mostly as being with motor functions, modulator and stabilizing functions. Addressing the contribution of muscles with different functions, Kornecki and Zschorlich (1994) found that while a manipulated object became progressively unstable the contribution of stabilizing muscles is increased while the activity of muscles with motor functions during that movement is decreased. In addition, it was shown that in the upper extremities the temporal coordination of muscles is phase shifted relative to each other when task demands are changing. This resulted in an advanced onset of stabilizing muscles relative to force producing muscles (Kornecki et al., 2001). These adjustments to changing biomechanical task demands arise, since muscles can cause destabilizing or counterproductive movements at some joints when they are contracted to create desired movements at other joints (Kornecki et al., 2001). In order to control the forces produced by muscles during movement only a certain fraction of the possibly available mobility of the motor system is needed at a given time (Kornecki 1992). According to Bernstein (1967) these adjustments can be interpreted

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as decreasing the inherent complexity in coordinated movements by the central nervous system by imposing suitable constraints to the motor system (Soechting and Lacquaniti 1989). These constraints were found in movements of the lower limbs (Nashner and McCollum 1985) or locomotion (Shik and Orlovsky 1976). Additionally, fixed contractile patterns were shown in postural stabilization (Nashner 1977). Conversely, McGill et al., (2003) pointed out that highly coordinated muscle activation patterns involving many muscles must change in dependence of the task to ensure (spine) stability. Consequently, the output of the motor system results from variable and task-dependent neuromuscular strategies which allow for specific solutions for the posed motor problem.

However, little is known about the underlying neuromechanical principles that govern patterns of muscle activation during movement performance (see also Ting 2007). Addressing the problem of the control of force exertion on an external object during multi-joint tasks, Kornecki (2001) suggested that the equilibrium state of the external object and not its mobility influences the forces produced by individual muscles. Together, the question is raised of whether the solution to the problem of controlling different DoF leads to specific changes in the temporal coordination patterns and gain of selected muscles when influenced by an external system of variable Degrees of Freedom.

The study was carried out to assess the form of the solution to redundancy by imposing different mechanical task constraints during performance of a multi-joint task with the lower extremities. The temporal characteristics of different muscles according to the change in equilibrium state of an external object are assessed by the use of a cross correlation function (CCF). It was hypothesized that the investigated muscles show an individual adjustment to changes in external mechanical features of the movement.

2. Materials and methods

The experiment involved 21 healthy male students of physical education (age 25.5 ± 1.2 SD, height $182.8 \text{ cm} \pm 5.3$ SD, weight $80.5 \text{ kg} \pm 8.6$ SD). The subjects received written and oral information about the procedures of the experiments before giving their written consent. The subjects were asked to produce their maximum of force on an external object during a unilateral knee extension with the right leg, with the object being influenced by varying Degrees of Freedom (Fig. 1).

2.1. General setup

The apparatus used is shown in Fig. 1 and from now on will be referred to as movement sled. The sled consisted of movable parts attached to a rigid frame which causes guided movements. The movable part contained a foot plate (Pos. 1 in Fig. 1) which could be varied in order to create changing stabilizing conditions. We tested in three different task conditions. In the first condition, the sled was in a fixed configuration with a rigid footplate where the subjects were exposed to 0 DoF. The second condition was in an unhitched configuration where the subjects were exposed to 1 DoF using the rigid foot plate. This condition permitted for movements along with the knee extension (z -direction in Fig. 1). The third condition allowed for 3 DoF. This condition comprised horizontal, rotational and translational movements which allowed for inversion/eversion and dorsal-/plantarflexion of the ankle. For this purpose an unstable footplate was employed, which was attached to the sled by a ball joint.

The movement sled moved on an inclined plane ($\alpha = 30^\circ$) with a mass of the movable parts of 18.2 kg. In order to produce force the subjects had to move a counterweight (Pos. 2 in Fig. 1) of 37.5 kg.

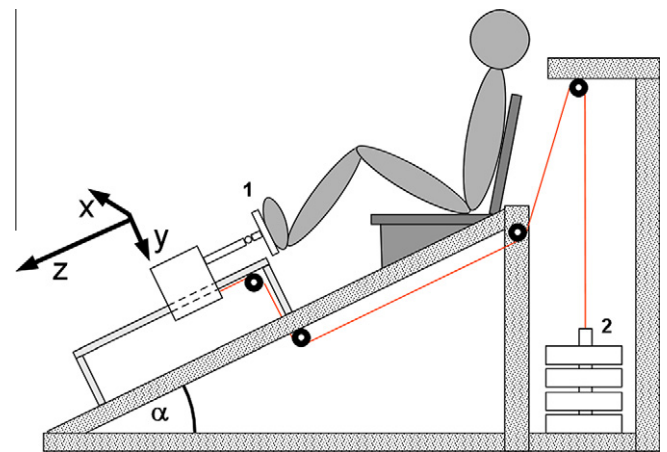


Fig. 1. General view of the movement sled. “ α ” (alpha) describes the angle of the inclined plane of the movement sled. “1” is the movable part of the sled with changeable footplates. In a fixed condition with a rigid foot plate the subjects were exposed to 0 DoF, in an unhitched condition with the rigid foot plate subjects were exposed to 1 DoF and with an unstable footplate subjects were exposed to 3 DoF. “2” refers to the position of the counterweight in order to produce resistance. The weights were 37.5 kg and were connected to the movable parts using a steel cable and pulleys. Defined movement dimensions are indicated.

2.2. Test procedure

Every subject had to complete five trials in each of the three conditions (0, 1 and 3 DoF). For each trial the initial sitting position and the position of the limb (right leg) was checked so that each subject started with a knee and ankle angle of 90° . The correct limb arrangement on the sled was checked by the use of a goniometer. The seat was adjustable to the height of the subjects which secured intersubject comparability. A 5 min warm-up period on a treadmill (at 6 km/h) had to be completed by each subject in advance of the test trials. During the data acquisition period a rest of at least 25 s between trials was given in order to reduce potential fatiguing effects due to movement execution. The order of the test conditions (0, 1 or 3 DoF) was randomized.

2.3. Data acquisition

The electromyographic (EMG) data and the three force components (F_x , F_y , F_z) were recorded simultaneously in every trial. The force in movement direction (F_z) is referred to as directional force (Bober et al., 1982), while F_x and F_y are referred to as non-directional forces. Surface EMGs (SEMG) were recorded from vastus medialis (VM), gastrocnemius medialis (GM), tibialis anterior (TA) and peroneus longus (PL). These muscles were chosen for this study because of their essential role in controlling anterior-posterior and lateral-medial stabilization of the knee and ankle joint (Cheng et al., 2004; van Ingen Schenau et al., 1987) and their contribution to functional stability (Suda et al., 2009).

The force signals were recorded with a three-dimensional Kistler force-transducer (Type 9602, Switzerland, Winterthur) with an integrated charge amplifier in a range of 0–2000 N. The device was positioned directly behind the foot plate. The three force components and the directions are depicted in Fig. 1.

SEMG-recordings for each muscle were made in a bipolar configuration (Merletti and Hermens 2004) with an interelectrode distance of 3 cm using Ag–AgCl electrodes (Hellige, diameter: 6 mm) which were filled with electrode gel. After standard skin preparation the electrodes were attached to the skin in line with the direction of the muscle fibres. The placement of the electrode was in accordance with the SENIAM recommendations (Hermens et al., 2000). The reference electrode was placed in the middle of the

recording electrodes for each muscle. The EMG-signals were pre-amplified (factor 1000) and high-pass filtered at 10 Hz using a Butterworth Filter of second order (Zschorlich 1989; De Luca et al., 2010). The cross-talk between the signals of the investigated muscles and the electrode placement was checked by reference contractions of each muscle prior to data recording. EMG signals were synchronized with the force data using a trigger of 200 N of the directional force time-curve with a pre-trigger of 300 samples (150 ms) at a sampling rate of 2 kHz on a DAQ-Card (Type 6024E, National Instruments, Austin, Texas, USA). Custom written software was used for analysis.

2.4. Data analysis

2.4.1. EMG and force Data

The rectified and smoothed (moving average smoothing filter with a time window of 25 ms) EMG signals were cut off from the first positive rise from baseline value (mean + 2SD) to evaluate the activity patterns of the investigated muscles. This approach ensured an individual assessment of the activity of each muscle in every single trial. The end of the movement was defined when the corresponding force-time curve (Fz) fell below 10 N after

reaching its maximal value (see Fig. 2). The integral of the filtered, rectified and smoothed EMG signal from movement onset to movement end was then calculated. The chosen time window had a variable length according to movement performance and thus represents the relevant phase of the stabilizing movement. The absolute peak values of all force components were used for evaluation of the maximal produced force during movement realisation.

2.4.2. Cross-correlation analysis

The cross-correlation function (CCF) was used to determine the temporal coordination between the selected muscles. The CCF was calculated from the data of single trials, the EMG being filtered, smoothed and full wave rectified and force data being smoothed. The CCF was calculated according to the formula (Person and Mislin 1964):

$$\varphi_{xy}(\tau) = \frac{1}{T} \int_0^T y(t) \cdot x(t + \tau) \cdot dt$$

The data were correlated in both directions, back- and forth, in order to evaluate positive and negative time shifts. Thus the corresponding cross-correlogram contained positive and negative x-values (Fig. 4a and b). The peak of the cross-correlogram was used for

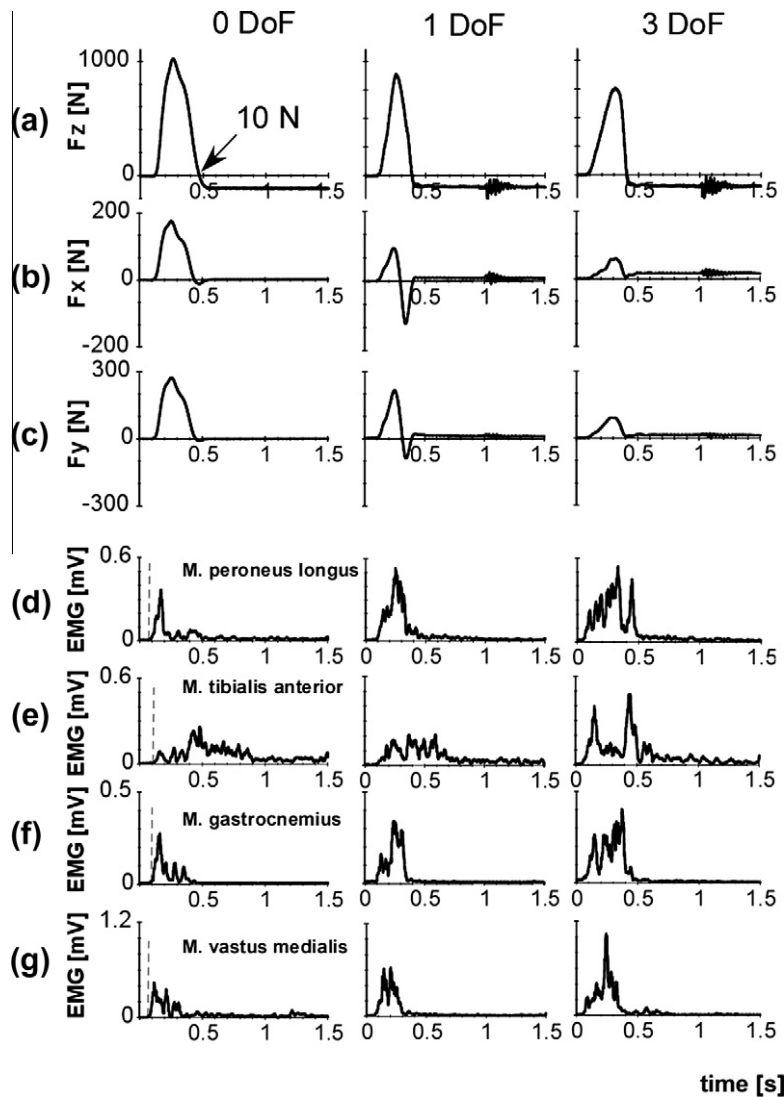


Fig. 2. (a–g) Representative data of a single subject during movement performance influenced by environmental conditions. (a) Directional force Fz; 10 N indicate the defined end of the movement; (b) Non-directional force component Fx; (c) non-directional force component Fy; (d–g) EMG profiles of the investigated muscles. The dotted lines show the defined onsets of activity (see methods). (Note the different scales used to provide the best fit to the data).

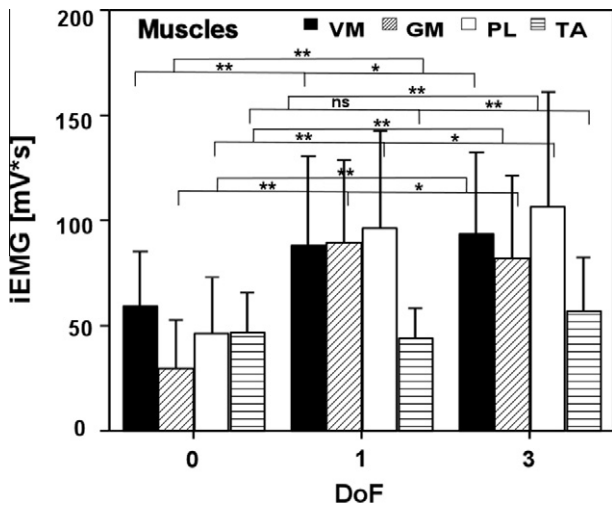


Fig. 3. Activities of the investigated muscles across all subjects in relation to changing environmental conditions. The calculated interval corresponds to the defined start and end points of the movement (see methods and Fig. 3). (Significances $p < 0.01$ **, $p < 0.001$ ***, ns = non significant).

assessing the time-lag τ of the two signals. Negative values indicated delayed phasing while positive values indicated advanced phasing. For assessment of the coordination patterns of the investigated muscles and components of force the EMG signals of VM always served as a reference signal. We have chosen this muscle endorsed by its function as a force producer and as initial component of the kinetic chain for the knee extension. This CCF approach allowed for a comparison of the signals among themselves. Regarding the analysis of two muscles a peak near the time of origin of the CCF indicates a high probability of co-activation of two muscles (Soechting and Lacquaniti 1989).

Paired t -test was used to assess statistical analysis. The reported results were considered significant for $p < 0.05$. Values in the text are given as mean \pm standard deviation (SD).

3. Results

3.1. Effect of external conditions

The principles of neuromuscular system adjustments were examined by quantifying changes in activation patterns between selected muscles of the lower limb. Electromyographic activity was recorded during execution of a multi-joint task of the right leg, which was subject to altered environmental conditions. A measure of muscle activity was obtained by considering the relevant time during force production and the temporal activity patterns were calculated by use of the cross-correlation function. Fig. 2a–g shows force and EMG profiles of a representative single subject.

As the DoF increased the peak directional force (F_z) decreased significantly ($p < 0.05$ for all conditions) (Fig. 2a, Table 1). In addition, the non-directional forces F_x and F_y also showed a decrease with increasing DoF. Significant differences emerged between 0 and 3 as well as between 1 and 3 DoF (Table 1). Because the supporting surface became progressively unstable, the decrease in F_x and F_y suggests an active stabilization process and not simply a passive consequence of actual motion.

3.2. Neuromuscular system adjustments

As the number of DoF changed from 0 to 1 DoF and 1 to 3 DoF, respectively, we observed an increase in the EMG impulse by the ankle stabilizing muscle PL (Table 2, Fig. 3). However, the TA

showed no difference between 0 and 1 DoF ($p = 0.182$), but its EMG impulse increased significantly in the 3 DoF condition relatively to the 0 and 1 DoF condition ($p < 0.001$, respectively). Simultaneously, the VM also increased its activity with the number of DoF ($p < 0.05$ for all conditions). The GM EMG activity showed a significant augmentation between static (0 DoF) and dynamic (1 and 3 DoF) conditions ($p < 0.001$), respectively. However, when we compared within dynamic conditions (1 and 3 DoF) we found that the GM decreased ($p = 0.013$).

The EMG-profiles (Fig. 2d–g) of the investigated muscles revealed patterns of co-activation in all of the investigated muscles since the ballistic nature of the task caused a temporal close activation. In order to examine whether a fixed set of movement patterns can account for the data in each condition, cross-correlations were computed between the activities of the muscles and the forces (see methods; Fig. 4a and b).

We computed maximum CCF values for each subject and compared the averages for the groups in the different task conditions (Table 3, Fig. 5). During movement realisation the CCF revealed positive values for all correlated pairs indicating a preceding activity of the vastus medialis muscle. This is not surprising since this muscle is part of the quadriceps femoris, the extensor of the knee joint and thus responsible for movement initiation. As the number of DoF increased the CCF revealed a significant progressive decrease of τ for all three force components ($p < 0.01$, respectively, Fig. 5).

As shown in Fig. 5e we observed no significant differences for the TA muscle between the conditions (0–1 DoF: $p = 0.081$, 0–3 DoF: $p = 0.109$, 1–3 DoF: $p = 0.791$). Nevertheless, it is noticeable that the TA exhibited a wide range of variation across all conditions (Table 3).

Furthermore, regarding the change in PL the 3 DoF condition revealed an advanced phasing compared to 0 ($p = 0.128$) and 1 DoF ($p = 0.008$), indicating patterns of enhanced co-activation. In contrast, the biarticular GM showed equal delays between 1 and 3 DoF ($p = 0.867$), but disclosed a significant shortening of τ in the 0 DoF condition resulting in a 31 ms (compared to 1 DoF, $p < 0.001$), respectively, 34 ms (compared to 3 DoF, $p = 0.003$) earlier onset of muscle activity (see Table 3). Nevertheless, the overall sequence of muscle recruitment (order: VM–PL–GM–TA) did not change.

4. Discussion

The study focused on the influence of varying Degrees of Freedom on movement coordination and stabilization. We performed a quantitative analysis of muscle activation patterns between selected muscles of the lower limb and components of force as environmental conditions changed. We hypothesized that the muscles show individual adjustments according to the change in external mechanical features of the movement. Based on our results the hypothesis is confirmed.

Consequently, the variations in the amount of exerted force and force onset occurred due to changes in the dynamic aspects and increased mobility of the external mechanical environment. Corresponding, we observed differences in the temporal activation patterns and in the gain of muscle activity. The changes in coordination patterns among the muscles in relation to the conditions point at a dependency on the status of the external system.

4.1. Changes in force output

The decrease in maximal directional force (F_z) during movement with increasing DoF (see Fig. 2a) is a consequence of the altered mechanical conditions of the external system. This finding is consistent with previous studies carried out on the wrist

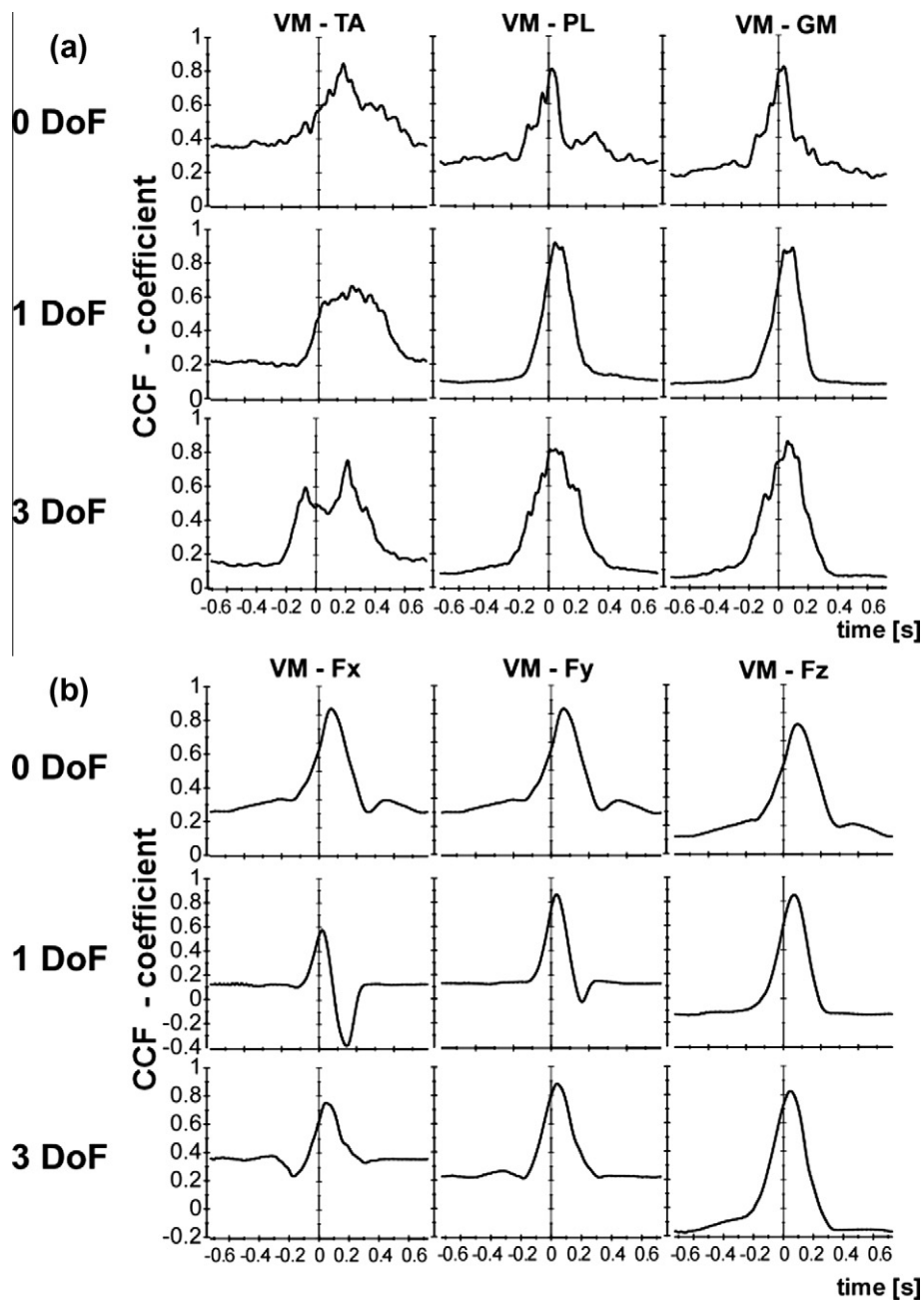


Fig. 4. (a and b) Cross-correlograms of the tested conditions between the activity of vastus medialis and tibialis anterior (VM-TA), peroneus longus (VM-PL), gastrocnemius medialis (VM-GM), Fy (VM-Fy), Fx (VM-Fx) and Fz (VM-Fz). Note the slightly different scales in a and b to provide the best fit to the data. The Data are from the same subject and trials as in Fig. 2.

Table 1
Mean and standard deviations of maximal forces produced during movement realisation. P values assessed by student's *t*-test.

Condition	Fz [N] Mean ± SD	Fx [N] Mean ± SD	Fy [N] Mean ± SD
0 DoF	969.2 ± 300	101.5 ± 75	160.4 ± 107
1 DoF	924.4 ± 205	114.8 ± 47	150.5 ± 77
3 DoF	721.7 ± 174	78.3 ± 31	107.8 ± 42
P 0–1 DoF	0.032	0.10	0.316
P 0–3 DoF	<0.001	0.002	<0.001
P 1–3 DoF	<0.001	<0.001	<0.001

Table 2
Mean and standard deviations of muscle activities produced during movement realisation, P values assessed by student's *t*-test.

Condition	TA [mVs] Mean ± SD	PL [mVs] Mean ± SD	GM [mVs] Mean ± SD	VM [mVs] Mean ± SD
0 DoF	46.84 ± 18.99	46.23 ± 27.6	29.7 ± 22.79	59.34 ± 25.68
1 DoF	44.03 ± 14.31	96.03 ± 46.68	89.32 ± 39.40	87.89 ± 42.61
3 DoF	56.71 ± 25.72	106.3 ± 54.86	81.97 ± 39.39	93.61 ± 38.51
P 0–1 DoF	0.182	<0.001	<0.001	<0.001
P 0–3 DoF	<0.001	<0.001	<0.001	<0.001
P 1–3 DoF	<0.001	0.011	0.013	0.043

(Kornecki and Zschorlich 1994; Bober et al., 1982). The fact that the maximum of force is produced in the fixed (0 DoF) condition and

subsequently decreased as the mechanical environment became progressively unstable suggests that the loss of force represents a

Table 3

τ values calculated by the CCF for principal components of force (top) and the investigated muscles (bottom). P values assessed by student's *t*-test.

Condition	Fz	Fx	Fy
	Mean \pm SD	Mean \pm SD	Mean \pm SD
0 DoF	96.8 \pm 42	112.8 \pm 61.1	145.9 \pm 126.4
1 DoF	63.9 \pm 26	75.4 \pm 60.4	84.9 \pm 118.5
3 DoF	35.1 \pm 51	40.0 \pm 126.4	46.3 \pm 77.6
P 0–1 DoF	<0.001	<0.001	0.001
P 0–3 DoF	<0.001	<0.001	<0.001
P 1–3 DoF	<0.001	0.009	0.001
	TA	PL	GM
0 DoF	142.9 \pm 196.3	30.7 \pm 64.6	34.2 \pm 56.9
1 DoF	195.5 \pm 229.7	36.2 \pm 50.3	63.5 \pm 57.2
3 DoF	187.7 \pm 213.7	16.6 \pm 65.2	64.6 \pm 73.4
P 0–1 DoF	0.081	0.502	<0.001
P 0–3 DoF	0.109	0.128	0.003
P 1–3 DoF	0.791	0.008	0.867

function of the growing instability. These differences were tested with student's *t*-test and turned out to be statistically significant at $p < 0.05$. Bober et al., (1982) attributed the diminishment of force to the process of muscle coordination. Thus, the changes of force magnitude can be regarded as a manifestation of neuro-muscular-skeletal system adaptation to unstable conditions of muscular force transmission to the mechanical environment (Kornecki 1992). We also observed a significant decrease in non-directional force components (Fx, Fy) with increasing external DoF. This find-

ing extends previous studies and suggests that the neuro-muscular system adjusts the overall force output in dependence of the mechanical environment.

4.2. Altered muscle activation patterns due to changing environmental conditions

The documented differences in the muscle activation patterns reflect responses to the altered external conditions in order to specifically control the muscle power. The distinct results of the activity patterns shown here might indicate a reconfiguration (see also Hogan 1985) in order to selectively react on effects of altered environmental conditions induced by movement realisation.

The decrease in the force output was accompanied by changes in the temporal coordination patterns. All three components of force showed a systematically decline in τ while the external DoF increased (Fig. 5, Table 3). This might indicate a preparation of the neuromuscular system for setting a more favourable condition for rapid force development by the muscles (Cavanagh and Komi 1979). Additionally, the control scheme of the muscles was characterized by an individual state-dependency accompanied by specific delays of motor reaction time between the muscles. More specifically, the ankle stabilizing PL and the VM always led the instant of force application for all three force components, whereas the TA always lagged the force onset (Fig. 5, Table 3). The magnitude of temporal shift depended on the condition and the muscle indicating selective neuromuscular system adjustments to the task.

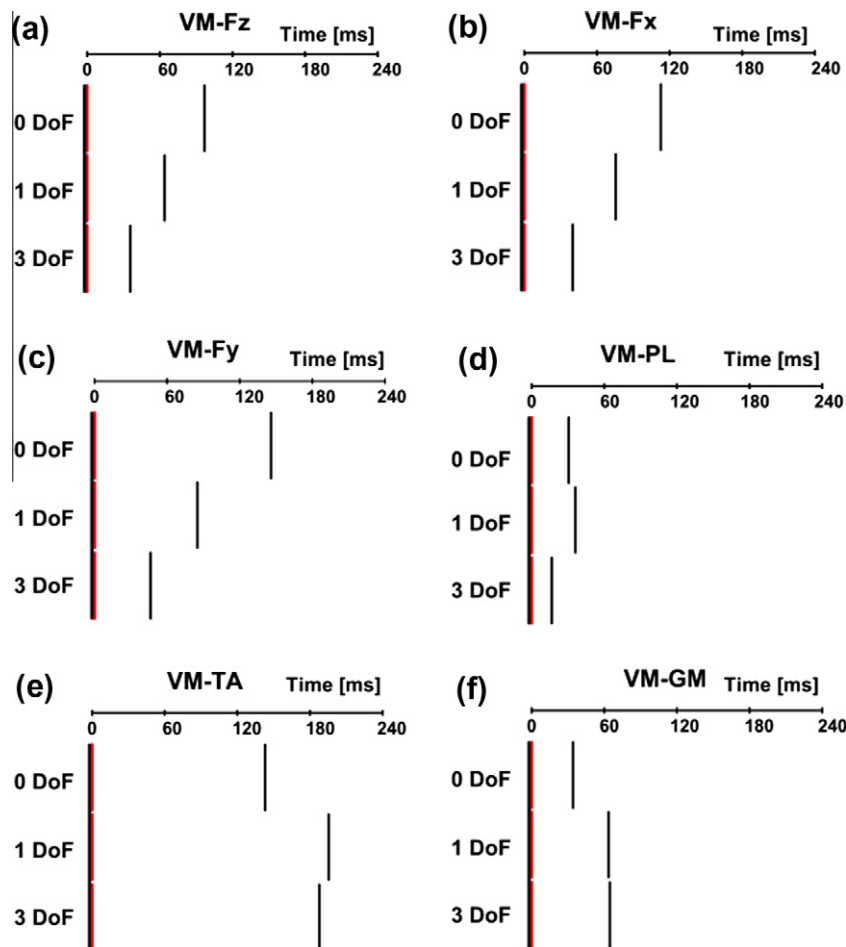


Fig. 5. (a–f) Values for τ for all correlated pairs of data. Data were averaged across all subjects. For all components of force a decline of τ could be shown. The muscles showed an individual behaviour depending on the function but the overall sequence did not change.

Therefore, the results indicate distinct control patterns between muscles with different functions during the movement. In case of PL the shortest delays for all three force components occurred in the 3 DoF condition underlining the functional significance of the muscle for the functional stabilizing process (see also Suda et al., 2009). The early onset of the PL and the late onset of the TA might allow, especially in the unstable condition, for an accurate and rapid control of muscle force. Furthermore, the GM showed patterns, which depended on the dynamic aspects of the movement, reflected in similar delayed onsets of activity in the dynamic conditions (1 and 3 DoF) as compared to the static 0 DoF condition. This sequence is in accordance to studies on the wrist (Kornecki et al., 2001) confirming the idea that not the mobility of the object being set in motion influences the forces produced by individual muscles. Especially multiarticular muscles like the GM bear multiple functions since they operate at several joints and thus develop different torques at the spanned joints simultaneously (Kornecki et al., 2001; van Ingen Schenau et al., 1987). In this context the delayed onset of GM in the dynamic conditions accompanied by a significant drop of activity between 1 and 3 DoF is an expression of necessary means in order to avoid a destabilizing influence of an inordinate activity. Therefore, it is suggested that the GM is more involved in the actual execution of the task. In the case of the TA no significant differences in τ emerged between conditions. This suggests that the timing of muscle activation was not different between these conditions giving the muscle a modulator function in the stabilization process. Goulart and Valls-Solé (1999) stated that stabilizing muscles but not prime movers are characterized by a great variability in the latency and amplitude with changing conditions. Consistent with that view the change of the mechanical conditions of the movement in this study did not change activity patterns of the TA significantly but revealed a high variability in onset latency. Furthermore, the relative delay of the TA left all postural adjustments to other muscles which in turn might have caused the observed advanced phasing of the PL in the 3 DoF condition (Goulart and Valls-Solé 1999).

As described earlier, the sequence of muscle activation (Fig. 5) always originated with the VM muscle. Additionally, this muscle showed the greatest amount of EMG activity which was further increased with increasing DoF (see Fig. 3 and Table 2). Thus, this muscle may be considered to be a major force producer during the movement and also contributes as knee stabilizer. Consequently, the VM contributes to both motor functions as well as stabilizing functions in this multi-joint task. Therefore, the early recruitment of the VM might contribute to an early stabilization of the knee before knee extension as it was suggested in the sit-to-stance manoeuvre (Cheng et al., 2004; Khemlani et al., 1999). This assumption is consistent with Bober et al., (1982), who suggested that in order to produce force by the powerful muscles with motor functions the biokinematic chain must be stable. This necessity is reflected in the increase of activity and the early onset of the knee stabilizing VM and the ankle stabilizing PL muscle and thus can be taken as expression of necessary adjustments by the neuro-muscular system to the environmental change. Further support for this idea comes from the fact that in our study the ankle stabilizing PL was recruited earlier than the movement producing GM, with the greatest difference occurring in the 3 DoF condition.

Thus, the effective transmission of force to the surrounding was achieved by enhanced co-operation between knee (VM) and ankle (PL) muscles (Hwang and Abraham 2001), and a simultaneous earlier onset of the PL relative to the GM under conditions of increasing instability. These results also agree with the finding that the wrist stabilizing muscles were activated prior to movement producing muscles during a similar task involving the upper extremity (Kornecki et al., 2001; Kornecki and Zschorlich 1994).

4.3. Functional implications

The adjustments in coordinative patterns across muscles influenced by changing mechanical conditions of the task are consistent with recent observations by Haddad et al., (2006) and Alibiglou et al., (2009). In our study, the existence of temporal differences between the selected muscles indicates that their recruitment is phase shifted relative to each other. Consequently, we suggest that specific temporal activation patterns in relation to the properties of the task were employed, but the overall sequence of muscle recruitment was preserved. The latter fact implicates that a centrally programmed sequence of activation is responsible (Goulart and Valls-Solé 1999). Furthermore, the changes in the EMG impulses indicates that the muscles are scaled in their contribution to individual components of the movement (stabilization or motor related) what in turn allows for a graded control of muscle power to meet stabilization demands (Da Fonseca et al., 2006). Therefore, the present results indicate that during movement realisation the motor apparatus just needs a certain fraction of its mobility. By putting variable active muscular constraints to the joints the stabilizing process is adjusted depending on the requirements of the task (see also Kornecki and Zschorlich 1994). Therefore, the coordinative interaction of stabilization and intended movement depends on the mechanical constraints of the movement (Rogers and Pai 1993). As pointed out by Hogan (1985) apparent redundancies or excess DoF increase the extent to which the mobility of the limb can be modulated. The control of movement is simplified by imposing neural constraints in order to reduce the number of DoF of the musculo-skeletal system.

5. Conclusions

Like it was shown in this study the intermuscular recruitment patterns of the investigated muscles may differ according to the function of the muscle and the condition of environment. Thus, the change in environmental conditions can influence the muscle coordination patterns of a ballistic force production task of the lower limb substantially.

The change in coordination strategies also caused changes in forces produced by the motor system, the forces acting on the environment being the final result of these strategies of the motor system. Finally, this study revealed that the functional stabilization process in the ankle joint is a manifestation of movement coordination.

The findings of this study further the understanding of the functional stabilization process and clarify the role of specific muscles during that process. This information might be useful for future studies and to detect potential changes that occur due to injury.

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The effect of increasing external degrees of freedom on force production and neuromuscular stabilisation

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Abstract

Evidence suggests that during interaction with different environmental dynamics the necessity to stabilise the involved joints leads to altered efficiency in force transmission to the surroundings and a specific orchestration of motor control strategies. However, little is known about the modalities of the changes associated with altered environmental dynamics. In 29 healthy participants, electromyographic (EMG) signals from four muscles of the right leg (*M. peroneus longus*, *M. tibialis anterior*, *M. vastus medialis*, *M. gastrocnemius medialis*) and three dimensions of force (F_x , F_y , F_z) were recorded. The participants were to exert force against an external object by performing a unilateral leg extension task with the task being influenced by either 0, 1 or 3 mechanical degrees of freedom. We hypothesised that the ankle stabilising muscles would increase their activities with increasing degrees of freedom (DoF), and that increasing external degrees of freedom results in decreased muscle force exerted during the movement task. The progressive change in the type of mechanical interaction from stable to unstable caused a loss of the ability to apply force in movement direction (F_z) which was accompanied by a reduction of F_y and F_x force dimensions. These reductions corresponded to maximum losses of 23% for F_z , 33% for F_x and 41% for F_y in the three degrees of freedom condition (all $P < 0.001$). Next, the individual muscles showed specific tuning effects, depending on the type of mechanical interaction. Our results suggest that the loss of the ability to exert force effectively against the external object is due to the neuromuscular stabilisation process of the involved joints. The change of the degrees of freedom conditions allowed for assessment of movement- or stabilisation-related adjustments of the motor system.

Keywords: ankle stabilisation, DoF, neuromuscular

1. Introduction

Human locomotion is not solely part in our everyday activities it is also an important component of our physical activities regardless of a competitive or recreational character. During sports and exercise the human motor system has to provide dynamic stability of the motor apparatus (Wikstrom, Tillmann, & Borsa, 2005) when facing, for example, altered joint loads due to landing after jumping (Gribble & Robinson, 2010) or during running at different velocities (Kyröläinen, Avela, & Komi, 2005). Consequently, the importance of ankle functionality as a part of the kinetic chain during human performances is well considered (Lephart, Pincivero, & Rozzi, 1998; Riemann & Lephart, 2002a, 2002b). Furthermore, when athletes experience changing biomechanical demands due to altered environmental conditions (e.g. running over compliant or uneven surfaces) an adequate control of joint stiffness is a prerequisite for controlling motion and avoiding potential injuries

(Stiles & Dixon, 2007). If the stability of movement is not secured by the mechanical environment the athlete is interacting with, the neuromuscular system has to provide that stability (Perreault, Chen, Trumbower, & Lewis, 2008). Accordingly, most sports create high demands on dynamic stability (Behm & Anderson, 2006) and thus (depending on the type of mechanical interaction) necessitate that the maintenance of stability (especially in unstable conditions) is provided by the neuromuscular system to varying extents. Furthermore, the necessary adjustments to task complexity might also affect the performance of an athlete (Norwood, Anseron, Gaetz, & Twist, 2007). Thus, the inherently greater instability of mechanical unstable movements challenges the neuromuscular system to a greater extent than under stable conditions (Behm & Anderson, 2006).

Kornecki (1992) highlighted that the cooperation of the nervous system with the motor system is an expression of necessary means facing the fact that in order to realise functional motor acts (e.g. exerting

force on an external object) only a small percentage of the potential mobility of the motor system is needed. Consequently, the process of functional stabilisation of joints was identified as consisting of the active restriction of unused degrees of freedom of the motor system (Kornecki, 1992). Therefore, during interaction with an external object, the coordinated use of muscles is fundamental (Herzog, 2000), especially when the mechanical properties of the object and hence the biomechanical demands of the task are changing. Accordingly, Kornecki and Zschorlich (1994) showed in the upper extremities that the efficiency of force transmission of a man-object-system decreases to around 80% if the object requires stabilisation. These findings are supported by Seo & Armstrong (2009), who indicated that the effect of stable as compared with unstable interactions is expressed by leverage-differences. Accordingly, leverage describes the ability to apply force in other than the movement direction for assistance of force production in the movement direction (Seo & Armstrong, 2009). Thus, the decrease in force output found in the mentioned studies might be linked to the reduction of F_y and F_x force output, which might reduce external moments about the joints (Seo & Armstrong, 2009). Furthermore, Aagaard, Simonsen, Andersen, Magnusson, & Dyhre-Poulsen (2002) highlighted that the rate of force development is a major determinant of maximal force during rapid movements. Changes in that parameter might have functional consequences for total force output (Aagaard et al., 2002), possibly affecting neuromuscular control strategies. Additionally, Kornecki et al. (2001) showed that the contribution of the stabilising muscles is increased and the activity of muscles with motor function is decreased while external objects become progressively unstable.

Accordingly, recent observations from our laboratory showed that the process of joint stabilisation is a manifestation of movement coordination also affecting the ability to exert force on the mechanical environment (Wuebbenhorst & Zschorlich, 2011). These coordinative processes were also shown to alter neural control mechanisms (Holl & Zschorlich, 2011). Taken together, the mentioned studies provided evidence that substantial changes in orchestration of movement control occur, also affecting the functional consequences of movement when biomechanical properties of the task are altered. However, little is known about the underlying mechanisms and the modality of the changes related to the process of functional joint stabilisation. Therefore, the aim of this study is to present further insight into the stabilisation process, the mechanisms used to achieve dynamic stability and its functional consequences. The employed method is conceptually based on the change of environmental conditions (and thus task demands)

during performance of a multijoint task involving the ankle and knee. In order to gain insight into movement- and stabilisation-related adjustments by the motor system we exposed our participants to different mechanical degrees of freedom during interaction with an external object (Holl & Zschorlich, 2011).

Based on previous studies that focused on the influence of differing degrees of freedom during movements of the upper extremities (Kornecki & Zschorlich, 1994; Kornecki, Keibel, & Siemienski, 2001) we hypothesize that: (1) the ankle stabilising muscles would increase their activities with increasing degrees of freedom, and (2) with increasing external degrees of freedoms the resultant muscle force exerted during the movement task decreases.

2. Methods

2.1. Participants

The experiment involved 29 healthy male students of physical education (age 26.3 ± 1.4 years, height 181.3 ± 6.1 cm, weight 79.8 ± 8.6 kg). All participants were naive to the experimental procedures. The study was conducted in accordance with the Declaration of Helsinki, approved by the local ethics committee. Prior to participating in the study, individuals read and signed a written consent form. No subject had a history of chronic ankle ligament or other lower extremities injuries. No subject presented orthopaedic disorders or shortening of the Achilles tendons.

2.2. Apparatus

The experimental approach describes a closed-kinetic chain (see also Escarmilla et al., 1998). The experiments were conducted on a test apparatus, which will be referred to as the movement sled (Figure 1). The sled moves on an inclined plane ($\alpha = 30^\circ$) with the movable parts attached to independently adjustable counterweights (Position 2 in Figure 1). For these experiments, the load was set to 37.5 kg, which described the maximum for the apparatus. Since it was shown by Bober, Kornecki, Lehr, and Zawadzki (1982) that with a reduced amount of required force the stabilisation is less important and coordination is paramount, we exposed the participants to more rather than less load in order to create more challenging conditions for the process of joint stabilisation.

The experiments involved three conditions, which differed by the employed number of degrees of freedom. Therefore, two different configurations of the sled and two different footplates (Position 1 in Figure 1) were used. The '0' degree of freedom condition was an isometric force production task with the sled being fixed in the movement direction (z -axis in Figure 1) by use of a rigid footplate.

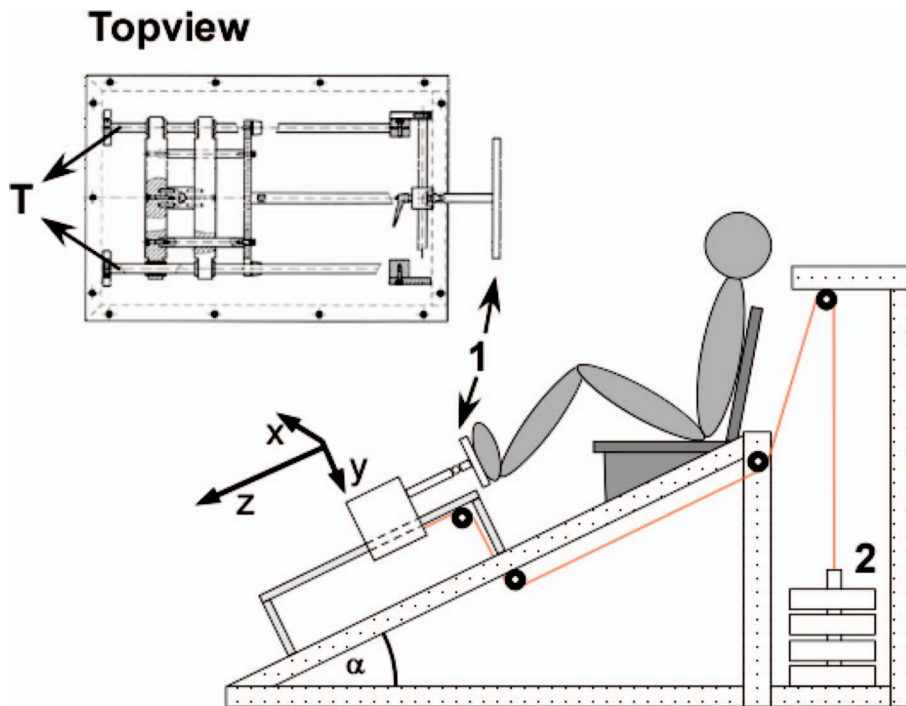


Figure 1. General view of the movement sled. ' α ' (alpha) describes the angle of the inclined plane of the movement sled. '1' is the movable part of the sled with changeable footplates. In a fixed condition with a rigid foot plate the participants were exposed to 0 DoF, in an unhitched condition with the rigid foot plate participants were exposed to 1 DoF and with an unstable footplate participants were exposed to 3 DoF. '2' refers to the position of the counterweight in order to produce resistance. The weights were 37.5 kg and were connected to the movable parts using a steel cable and pulleys. Defined movement dimensions are indicated. The top-view insertion outlines the construction of the movable parts of the sled. The sled runs on two guiding tubes ('T') and is connected with the foot plate and the counterweight.

During increased degrees-of-freedom trials the equilibrium state of the sled was changed by using an unhitched configuration of the sled. In this configuration we exposed the participants to either one degree of freedom (along the z -axis) by using the rigid footplate or to three degrees of freedom by using an unstable footplate, which was attached to the system by a ball joint. The three degrees of freedom comprised horizontal, rotational and translational movements, which allowed for inversion/eversion and dorsal/plantar flexion of the ankle. The comparison of the experimental conditions allows for identification of movement- and stabilisation-related adjustments by the motor system (Holl & Zschorlich, 2011).

The seat of the sled was adjustable in height to account for different anthropometric requirements of the participants. For standardisation purposes every movement started with a knee and ankle angle of 90° measured manually with a goniometer. In the three degrees of freedom condition, the position of the foot was controlled by aligning the ankle joint centre with the centre of rotation of the external foot plate.

2.3. Experimental design

The present experiments were carried out using a within-subject experimental design, the factors being

degrees of freedom-condition (DoF-condition 0, 1, 3), muscle (M. tibialis anterior [TA], M. peroneus longus [PL], M. gastrocnemius medialis [GM], M. vastus medialis [VM]), and force component (F_x , F_y , F_z). Each participant completed the three DoF-conditions, with five trials per condition. The participants were instructed to perform a maximal ballistic unilateral leg extension movement along the z -axis using their right leg. The unemployed left leg rested during the trials. Prior to each trial the participants were asked to position their foot on the support plate with no activity in the investigated muscles (checked by observation of EMG-time-signals with high resolution [± 0.1 mV]) and the force sensors were reset to zero. The movement started with no preceding countermovement as controlled by visual observation of task kinematics and by control of the directional force-time curves. For each randomised condition, five trials were recorded resulting in a total of 15 trials for each subject. Since most sports demand high velocities of motion (Willardson, 2004) this study employed a task that mimics the effort of naturally occurring high velocity movements.

2.4. Data recordings and processing

Surface EMG measurements were made using Ag/AgCl electrodes (Hellige baby electrodes, GE

Medical Systems, Germany) in a bipolar configuration. The electrode pairs were spaced about 4 cm apart on the skin along the longitudinal axis of the muscles and with the ground electrode placed between the recording electrodes for each muscle (Merletti & Hermens, 2004). A factor of 1000 was used for amplification of myopotentials. The skin surface was prepared according to the generally accepted principles (Basmajian, 1978). The measurement of the force components was carried out using a three-component force sensor with integrated electronics (KISTLER Type 9602, Switzerland, Winterthur). The force sensor was placed directly behind the footplate (Position 1 in Figure 1). Three components of force were recorded and defined according to Figure 1. The F_z -component will be referred to as 'directional force'. The total data collection time was 1.5 s with a sampling rate of 2 kHz and EMG signals were sampled simultaneously with the force data using a trigger of 200 N of the directional force time-curve with a pre-trigger of 300 samples (150 ms). The four muscles measured in this study were: the M. vastus medialis, the M. gastrocnemius medialis, the M. tibialis anterior and the M. peroneus longus. The M. gastrocnemius medialis and M. vastus medialis (as part of the quadriceps femoris muscle) serve as representatives for muscles with mainly motor functions and M. tibialis anterior and M. peroneus longus describe muscles with primarily stabilising functions (Wuebbenhorst & Zschorlich 2011). The force and EMG signals were digitised using a 12 bit A/D converter DAC-Card (Type 6024 E National Instruments, Austin, Texas, USA) and stored for further processing. Custom written software was used for analysis.

The EMG signals were filtered using a high-pass Butterworth filter (IIR) with a cut-off frequency of 10 Hz to eliminate artefacts of the raw signals (Zschorlich, 1989). Subsequently, the signals were rectified, smoothed (moving averaged smoothing filter 25 ms) and integrated (iEMG). For calculation of iEMG, the movement start was defined by the first positive rise of EMG of a given muscle from baseline value (mean + 2 standard deviations [SD] of the first 50 ms of recording) and the end of movement being defined when the directional force (F_z) fell under a value of 10 N on the descending limb of the force-time curve. This approach allowed for an individual assessment of a given muscle based on the particular performance of the task.

The identification of the maximum of each force component was made by the calculation of the peak values of the corresponding force-time curve. The rate of change of force development was assessed by differentiation of the directional force F_z . Peak values of the corresponding curve were used for estimation of this parameter.

As an example, synchronised courses of directional force with raw EMG records are shown in Figure 2 for one participant.

2.5. Data analysis

Data are presented as mean \pm one standard deviation (SD). Absolute values are presented in the figures and tables and are described frequently in the results section. Relative changes are expressed as a percentage of change ($\% \Delta$) followed by the parameter and the two conditions that were compared (example: $\% \Delta F_{z0-3}$ for comparison of F_z between 0 and 3 degrees of freedom). With respect to our experimental design we used a two way, within-participants ANOVA with repeated measures 'force component' (F_y , F_x , F_z) and 'DoF-condition' (0, 1, 3) for analysis of force data. Interaction effects

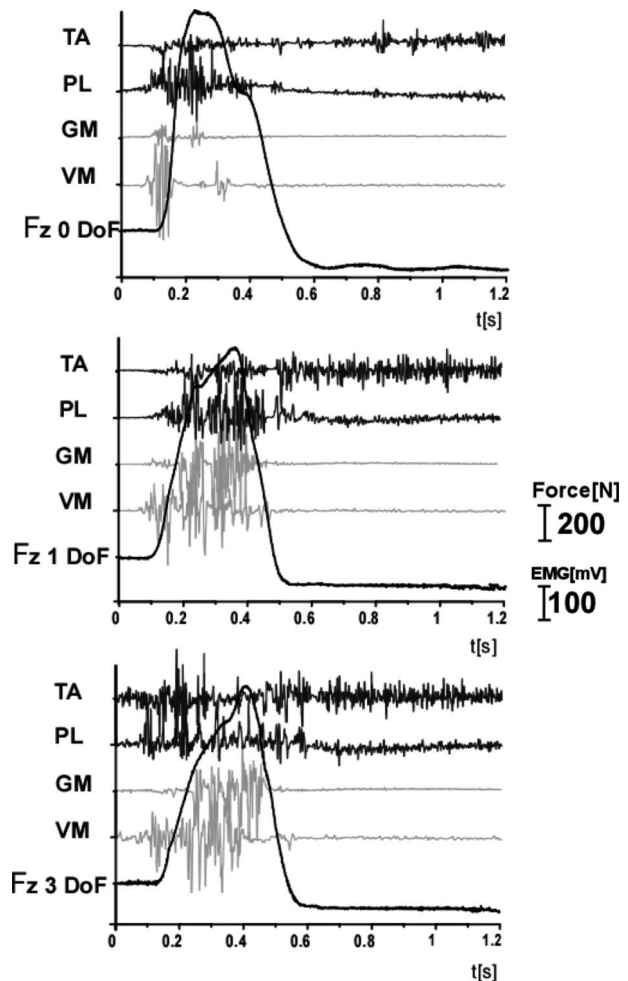


Figure 2. Time courses (of one individual participant) of the force in movement direction (F_z) and of the EMG signals (unfiltered raw data) of the investigated muscles of the lower extremities for 0, 1 and 3 DoF, respectively. The chosen muscles represent muscles with kinetic (GM and VM) or stabilising function (TA and PL) in this multi-joint movement.

for force component (F_z , F_y , F_x) \times DoF-condition (0, 1, 3) were evaluated. The rate of force development was analysed using a one-way ANOVA with DoF-condition as an independent factor. We subjected the data of measured muscular activity to a two-way repeated measures within-participants ANOVA (evaluated interaction: DoF-condition [0, 1, 3] \times muscle type [TA, GM, VM, PL]). In the case of significant main effects, post-hoc analysis was performed using Student's t-test to evaluate the significance of the differences between DoF-conditions.

Reported results are considered significant for P , at $\alpha=0.05$. The data were normally distributed (Pearson chi-square normality test). All statistical analyses were conducted using SPSS 19.0 (SPSS Inc., Chicago, Illinois, USA).

3. Results

The altered mechanical environmental conditions resulted in a decrease of force with increasing degrees of freedoms for all components of force (Figure 3, Table I). Our statistical analysis revealed a significant interaction effect for DoF-condition \times

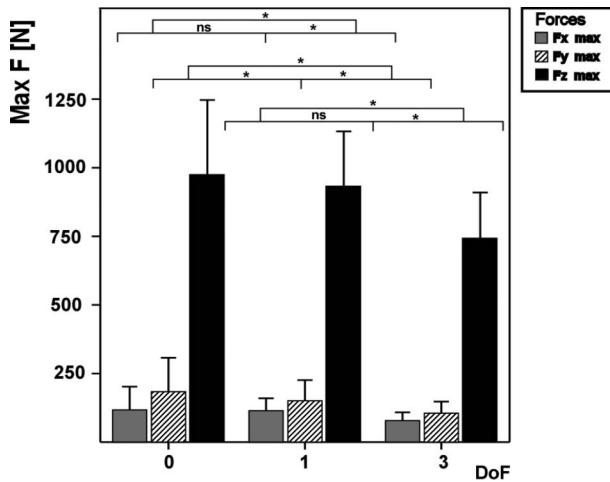


Figure 3. Maximum forces F in three movement dimensions (F_x =horizontal force, F_y =vertical force, F_z =resultant force in movement direction) with increasing DoFs. A significant interaction effect for the DoF condition \times force component was found ($P < 0.001$).

force component [$F(4) = 10.35$, $P < 0.001$] for this measure. The values of the directional force F_z reached a maximum in the 0 degree of freedom condition (975 N \pm 270 N) and subsequently decreased reaching 933 N (\pm 199 N) for one degree of freedom and 743 N (\pm 167 N) in the three degrees of freedom condition. Our post-hoc analysis showed that in the comparison of the particular conditions the relative losses of directional force were $\% \Delta F_{z_{0-1}} = -4.34\%$ ($P = 0.0979$), $\% \Delta F_{z_{0-3}} = -23.7\%$ and $\% \Delta F_{z_{1-3}} = -20.3\%$ ($P < 0.001$, respectively). The force components F_x and F_y also showed significant decreases depending upon the status of the external system. While increasing the degrees of freedoms the decrease of $\% \Delta F_{y_{0-1}}$ was -18.4% ($P = 0.017$), for $\% \Delta F_{y_{0-3}} = -41.6\%$ ($P = 0.0012$) and for $\% \Delta F_{y_{1-3}} = -28.4\%$ ($P = 0.0051$). For F_x the relative differences corresponded to $\% \Delta F_{x_{0-1}} = -2.63\%$ ($P = 0.955$), for $\% \Delta F_{x_{0-3}} = -33.5\%$ ($P = 0.0216$) and for $\% \Delta F_{x_{1-3}} = -31.7\%$ ($P = 0.0005$).

The different environmental dynamics also exhibited significant changes in the slope of force development (Figure 4, Table III). Thus, we found a significant main effect of DoF-condition for this measure [$F(2) = 3.127$; $P = 0.04$]. The post-hoc analysis clarified that with increasing degrees of freedom the slope declined significantly. For 0, 1 and 3 degrees of freedom the apparent changes calculated from the directional force F_z corresponded to $15,369 \text{ N} \cdot \text{s}^{-1}$ (± 8389), $13,330 \text{ N} \cdot \text{s}^{-1}$ (± 5051), and $11,837 \text{ N} \cdot \text{s}^{-1}$ (± 4277). Changes between 0 and 1 degrees of freedom were significant with $P = 0.002$, and for 0-3 and 1-3 significant with $P < 0.001$ (Table III).

The iEMGs of the investigated muscles showed non-uniform changes between the conditions (Figure 5, Table II) indicating selective gain scaling according to muscle function. This was supported by a significant interaction effect between DoF-condition and muscle [$F(6) = 18,135$, $P < 0.001$]. When subjected to our post-hoc analysis the M. tibialis anterior exhibited relative changes according to $\% \Delta TA_{0-1} = 6\%$ ($P = 0.17$), for $\% \Delta TA_{0-3} = 32\%$ and for $\% \Delta TA_{1-3} = 24\%$ ($P < 0.001$, respectively). For M. peroneus longus the comparison of the static and dynamic conditions revealed significant changes in activity resulting in relative differences of $\% \Delta PL_{0-1}$

Table I. Mean and standard deviations of maximal forces produced during movement realization. P values assessed by Student's t-test.

Condition	F_x [N] Mean \pm SD	F_y [N] Mean \pm SD	F_z [N] Mean \pm SD
0 DoF	117.90 \pm 85.46	184.51 \pm 123.22	975.34 \pm 270.52
1DoF	114.80 \pm 46.64	150.52 \pm 76.75	933.05 \pm 199.28
3 DoF	78.32 \pm 31.27	107.77 \pm 42.04	743.34 \pm 167.89
P 0-1 DoF	0.955	0.017	0.0979
P 0-3 DoF	0.0216	0.0012	<0.0001
P 1-3 DoF	0.0005	0.0051	<0.0001

= 123% and $\% \Delta \text{PL}_{0-3} = 132\%$ ($P < 0.001$, respectively). The comparison of dynamic conditions (1 and 3 degrees of freedom) revealed a value $\% \Delta \text{PL}_{1-3} = 3\%$, without reaching any statistical significance ($P = 0.23$).

Having regard to the muscles with mainly motor functions in this experiment it can be noted that both (*M. vastus medialis* & *M. gastrocnemius medialis*) increased significantly in comparison of the static (0 degree of freedom) and dynamic conditions (1 and 3 degrees of freedom), numerically reflected by $\% \Delta \text{VM}_{0-1} = 44\%$, $\% \Delta \text{VM}_{0-3} = 57\%$, $\% \Delta \text{GM}_{0-1} = 175\%$, and $\% \Delta \text{GM}_{0-3} = 159\%$ ($P < 0.001$, respectively). Additionally, the comparison of the dynamic

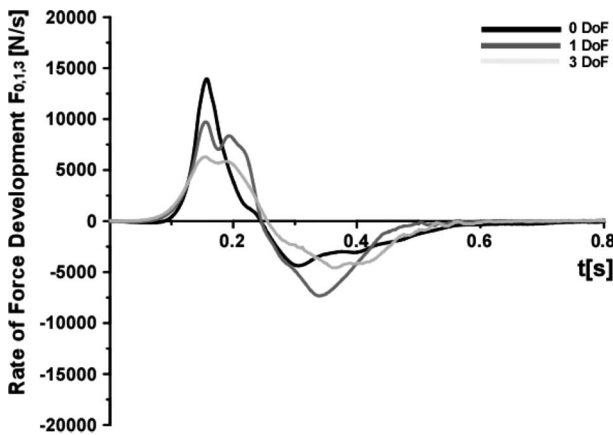


Figure 4. Mean rate of force development for all trials for resultant forces F_z . A significant main effect for DoF-condition (0, 1, 3) was found for this measure ($P = 0.04$).

conditions revealed a further increase in activity for *M. vastus medialis* ($\% \Delta \text{VM}_{1-3} = 9\%$, $P = 0.0325$) and a decrease for *M. gastrocnemius medialis* ($\% \Delta \text{GM}_{1-3} = -9\%$, $P = 0.033$).

4. Discussion

This study focused on the effects of increasing external degrees of freedom on ankle stabilisation strategies during a unilateral voluntary force production task. We hypothesised that increasing external degrees of freedom results in decreased force transmitted to the environment. This was confirmed by our data (Figure 3). Furthermore, we hypothesised that increasing external degrees of freedom results in altered regulation processes of the activities of stabilising muscles involved in the stabilisation process of the ankle complex. As shown in Figure 5, this hypothesis was also supported by the data.

This study was carried out based on the assumption that different support conditions cause context specific adjustments according to the biomechanical exigencies of the task. We found that the effect of increasing external degrees of freedom results in a drop of the effective force transmission to the surrounding as evidenced by a significant interaction effect for DoF-condition \times force component. This finding is in accordance with earlier studies of the wrist joint (Bober, Kornecki, Lehr, & Zawadzki, 1982; Kornecki & Zschorlich, 1994; Kornecki et al., 2001). Extending the results of previous studies we showed that all force components (F_y and F_x) as well as the rate of change of directional force development are diminished. The finding of altered rate of

Table II. Mean and standard deviations of muscle activities produced during movement realisation. A significant interaction effect for DoF-condition (0, 1, 3) and muscle type (GM, TA, PL, VM) was found ($P = 0.0004$). P -values in contrast analysis assessed by Student's t -test.

Condition	TA [mV · s] Mean \pm SD	PL [mV · s] Mean \pm SD	GM [mV · s] Mean \pm SD	VM [mV · s] Mean \pm SD
0 DoF	46.88 \pm 18.66	45.89 \pm 26.96	33.36 \pm 28.79	64.59 \pm 31.84
1 DoF	49.69 \pm 21.88	102.62 \pm 49.95	91.84 \pm 40.37	93.22 \pm 40.22
3 DoF	61.96 \pm 30.88	106.56 \pm 54.19	83.68 \pm 36.16	101.76 \pm 38.65
P 0-1 DoF	0.17	<0.0001	<0.0001	<0.0001
P 0-3 DoF	<0.0001	<0.0001	<0.0001	<0.0001
P 1-3 DoF	<0.0001	0.23	0.033	0.0325

Table III. Maximum of the rate of force development of the resultant force exerted on the external object with adjustable mobility DoF 0, 1, 3 in $\text{N} \cdot \text{s}^{-1}$. P -values assessed by Student's t -test.

	0 DoF [$\text{N} \cdot \text{s}^{-1}$]	1 DoF [$\text{N} \cdot \text{s}^{-1}$]	3 DoF [$\text{N} \cdot \text{s}^{-1}$]	Significance P		
				0/1 DoF	0/3 DoF	1/3 DoF
Mean	15369.15	13330.20	11837.43	0.002	<0.0001	<0.0001
SD	± 8389.05	± 5051.31	± 4277.01			

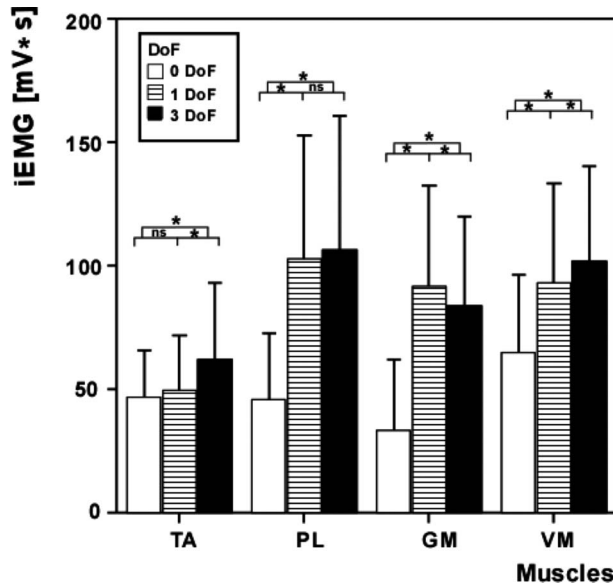


Figure 5. Mean iEMGs of the muscles with motor function (VM, GM) and stabilising function (PL, TA) engaged in setting in motion the external object with adjustable mobility. (Significances $P < 0.0001$ ‘*’, ns = non significant).

force development could be attributed to a strategy by the motor system based on minimised interference during the process of force propulsion on the man–object system in conditions with increased degrees of freedom. This view is supported given the fact that the type of mechanical interaction was a determining factor. Considering that the static 0 degree of freedom condition allowed for the maximum of force (Figure 3), the drop of force in the dynamic conditions represents a function of the growing instability. Thus, the force efficiency of the man–object system is decreased in conditions that employ increased degrees of freedom. Another study of upper limb joint stabilisation found that the directional force drops by approximately 20% when the external object requires stabilisation (Kornecki & Zschorlich, 1994). Our results therefore support these findings by approaching this value when comparing the 3 degrees of freedom condition to either 1 or 0 degrees of freedom. Further, F_x and F_y force components showed an enhanced loss of efficiency in dynamic conditions reaching a maximum of 33% for F_x and 41% for F_y in the 3 degrees of freedom condition. Thus, the mastering of a certain amount of degrees of freedoms greater than zero causes a loss of effective force transmission to the surrounding. Additionally this loss is sensitive to the status (and therefore to the biomechanical properties) of the external system. The results of the present study are therefore consistent with the idea that the reduction of the F_x and F_y force dimensions during unstable interactions serves to

reduce external moments at the joints (Seo & Armstrong, 2009). Accordingly, stable interactions are believed to provide leverage, the ability to apply force in other than the movement direction, which will assist the directional force production (Seo & Armstrong, 2009). Thus, the current findings are well in line with this view. Nevertheless, considering the nature of the present task (isometric vs. stable dynamic vs. unstable dynamic condition) the force–velocity relation of muscular force production might have contributed to the reported results. This relation might have affected the leg extension movement in comparison with the experimental conditions. However, this effect might be inherent to the neuromuscular stabilisation process itself, the reported effects on force transmission being the final result of these strategies of the motor system.

From a functional perspective, the loss of effective force transmission due to instability is crucial for a variety of human performances. For example, Anderson and Behm (2004) reported a loss of force of 60% during an isometric chest press task being influenced by instability. These effects have to be considered when using heavy weights (e.g. during rehabilitation programmes) under unstable conditions in order to activate high-threshold motor units (Behm & Anderson, 2006). Furthermore, the interaction with different environmental dynamics is a common approach in proprioceptive training and has been shown to produce improved neuromuscular activation (Cressey, West, Tiberio, Kraemer, & Maresh, 2007). Therefore, the results of the present study might be useful when designing programmes for rehabilitation or for improvement in athletic performance based on different mechanical conditions of the environment.

Considering possible mechanisms for the reduced ability to produce force Bober et al. (1982) suggested that the drop of resultant force rests on the physiology of muscle coordination. Thus, the decrease is associated with the necessity to stabilise the involved joints as the weak links in the system of force production. Referring to the process of stabilisation it has been suggested that the engagement of only a certain fraction of the potential mobility of the motor system is required (Kornecki & Zschorlich, 1994). By actively restraining the involved joints during the movement the posed problem of controlling excessive degrees of freedom is approached. Kornecki (1992) stated that this would ensure the blockade of some joints in order to achieve simultaneously coupled motions taking place in other joints. Accordingly, as was shown in this study, the functional stabilisation process of a multi-joint task is expressed in terms of a specific regulation of the stiffness properties of the involved joints as evidenced by the change of activity of the investigated muscles (Figure 5, Table II). It is

important to note that individual muscle contributions are not independent and could not be decoupled in the different conditions (McGowan, Kram, & Neptune, 2009). Nevertheless, the changes reflect the adjustments by the motor system to meet the altered mechanical demands of the experimental conditions. This study revealed that the changes in muscle activity resulted in a specific orchestration of activity (Figure 5). In particular, the *M. tibialis anterior* showed no difference between 0 and 1 degrees of freedom, but revealed a significant increase in comparison to 3 degrees of freedom. Further, the *M. peroneus longus* showed no change in comparison of the dynamic conditions (1 & 3 degrees of freedom) but showed significant lower EMG amplitudes in the static (0 degree of freedom) condition, respectively. These adjustments constitute a prerequisite for movement control, keeping in mind that in order to exert force with the powerful muscles with motor function, the biokinematic chain must be stable (Bober et al., 1982). This approach contains the stabilisation of the involved joints while creating the propulsive mechanical energy of the prime movers. In this study, this requirement was fulfilled by selectively increasing the activities of the stabilising muscles in unstable situations. However, it has to be noted that, as shown before (Wuebbenhorst & Zschorlich 2011) the *M. peroneus longus* and *M. tibialis anterior* revealed different responses to the changes in environmental dynamics. This is explained by different functional contributions of these muscles to the process of joint stabilisation.

Kornecki et al. (2001) previously hypothesised that while the external object becomes progressively unstable, the activity of the muscles with motor functions is attenuated by the central nervous system. Thus, the finding that the activity of the *M. vastus medialis* as a prime mover in this study increased between conditions is conflicting with this hypothesis. Therefore, we reasoned that the increase reflects a contribution to the stabilisation of the knee during the movement; giving this muscle motor as well as stabilising functions in this multi-joint task. This view is supported by McGowan et al. (2009) who indicated that a muscle's function is defined by its contribution to multiple tasks. Furthermore, this idea is well in compliance with observations by Anderson and Behm (2004) and Behm and Anderson (2006) who reported a loss of force due to instability, not to be associated with changes in EMG activity. In accordance with our assumption, the authors suggested that the motive force of the muscles was transferred into greater stabilising forces.

The *M. gastrocnemius medialis* is a biarticular muscle and thus exerts forces at the knee and ankle joint simultaneously (van Ingen-Schenau, Bobbert, & Rozendal, 1987). Therefore, this muscle increased its mechanical output between static and dynamic

conditions ($\% \Delta GM_{0-1}$ 175%; $\% \Delta GM_{0-3}$ 159%) in order to contribute to both ankle extension and knee flexion (Zajac, 1993). On account of this, we suggest that the drop of activity between the dynamic conditions ($\% \Delta GM_{1-3} = -9\%$) reflects a mechanism to diminish the destabilising potential of an inordinate activity in order to maintain the equilibrium (Nashner, 1976; Frank & Earl, 1990; Holl & Zschorlich, 2011).

The methods employed in the present study enabled us to study the effect of differing environmental dynamics. Hence, this approach extends previous studies by identification of motor and stabilisation related changes of the motor system and thus contains a potential for use in treatments of movement disorders or for design of programmes for rehabilitation after injury or for athletic performance improvements. Consequently, our results further the understanding of the mechanisms used to achieve dynamic stability involving multiple joints (see also Arampatzis, Karamanidis, & Madelml, 2008).

The present study is characterised by a homogeneous demographical structure and a relatively small sample size. Thus, a 'healthy worker' bias might have been introduced into the study. Consequently, there might be an underestimate of the effects of age, gender or physical status. However, considering comparable results under conditions of instability from studies involving both genders (Norwood et al., 2007) the results of the present experiments are well in line with this study. Nevertheless, bias-effects due to the design might have influenced certain result characteristics.

In conclusion, our investigation showed that changes in the mechanical environment of a task lead to specific adjustments of the process of joint stabilisation and thus for maintenance of dynamic stability. Accordingly, the specific changes in the activity of the investigated muscle suggest that coordinative patterns are selective in relation to the biomechanical exigencies of the task. As shown here, the selectivity of neuromuscular responses is dependent on muscle type and environmental conditions, respectively. It is suggested that suitable constraints are imposed onto the motor system in order to coordinate the functional stabilisation process of joints. Furthermore, the study demonstrates that the maximal forces that are applicable to the environment depend on the type of mechanical interaction, showing a decrease with increasing instability of the mechanical environment.

Conflict of interest statement

The authors have disclosed any financial and personal relationships with other people or organizations that could inappropriately influence their work.

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Interaction against different environmental dynamics during a leg extension task is controlled by temporal rather than amplitude scaling of muscular activity

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ABSTRACT

Force exertion against different mechanical environments can affect motor control strategies in order to account for the altered environmental dynamics and to maintain the ability to produce force. Here, we investigated the change of muscular activity of selected muscles of the lower extremities while the participants interacted with an external mechanical device of variable stability. Twenty-five healthy participants exerted force against the device by performing a unilateral ballistic leg extension task under 1 or 3 degrees of freedom (DoF). Directional force data and electromyographic responses from four leg muscles (TA, VM, GM, PL) were recorded. Muscle responses to the altered experimental conditions were analyzed by calculating time to peak electrical activity (TTP), peak electrical activity (PEA), slope of EMG-signal and muscle activity. It was found that neuromuscular system adjustments to the task are expressed mainly by temporal (TTP) rather than amplitude (PEA) scaling of muscular activity. This change was specific for the investigated muscles. Moreover, a selective increase of muscle activity occurred while increasing external DoF. This scheme was accompanied by a significant reduction of applicable force against the device in the unstable 3 DoF condition. The findings suggest that orchestration of movement control is linked to environmental dynamics also affecting the ability to produce force under dynamic conditions. The adjustments of the neuromuscular system are rather temporal in nature being consistent with the impulse timing hypothesis of motor control.

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1. Introduction

The well-coordinated use of muscles during all kinds of activities is known to be a prerequisite for motor control strategies facing the redundant nature of the neuro-musculo-skeletal system. Because of the redundancy, the interaction with various environmental conditions constitutes a requirement for an adaptive control of stiffness properties of the involved joints. These motor control strategies provide robustness to motor output variability and perturbations from the environment (Franklin et al., 2003). However, from biomechanical studies it is well known that neuromuscular control patterns change in response to different stability conditions and affect the forces transmitted to the surrounding (Bober et al., 1982; Kornecki and Zschorlich, 1994; Kornecki et al., 2001). We have recently identified distinct intermuscular temporal activation patterns and selective gain changes of the involved muscles as cardinal determinants of these neuromuscular adjustments of the stabilization process (Wuebbenhorst and

Zschorlich, 2011) as well as changes in neural control strategies (Holl and Zschorlich, 2011). These studies support the idea by Rancourt and Hogan (2009) that force exertion against the mechanical environment can destabilize motion and compromise on the ability to exert force. Franklin et al. (2003) distinguished control strategies of stable and unstable tasks pointing out that unstable tasks require muscular activation levels above and beyond that usually occurring with the movement. However, the increase in muscular activation due to stabilization often cannot be used for effective force transmission to the surrounding as evidenced by decreased peak forces (Bober et al., 1982; Wuebbenhorst and Zschorlich, 2011) or by reported differences for predicted joint stiffness from net joint torque necessary for producing the movement under unstable conditions (Franklin et al., 2003). Therefore, the instability created by the muscular force itself may limit the force magnitude (Rancourt and Hogan, 2001a,b). Additionally, the ability to coordinate muscle groups for producing rapid, goal-directed movements is indispensable and was shown to be a major factor for performance losses in rapid force production in the elderly (Barry et al., 2005). However, in order to stabilize movements under varying conditions the motor system uses stiffness regulation processes of the involved joints (Rancourt

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and Hogan, 2009). As shown by Hunter and Kearney (1982) stiffness properties scale with muscular activation level. Nevertheless, when stabilizing a joint, co-contraction of the agonist–antagonistic system provides a means for counteracting a destabilizing force and alters joint stiffness (Heitman et al., 2011). However, as shown by Milner et al. (1995) muscle activation capability can be reduced by 50% when antagonistic cocontraction is required. The latter fact highlights the importance of a proper coordination strategy.

Hence, we concluded that the stabilization process (and consequently the maintenance of the ability to exert force) is a manifestation of movement coordination (Wuebbenhorst and Zschorlich, 2011). Accordingly, the mentioned studies showed that the mechanical coupling of a limb and an external object causes neuromuscular system adjustments to the actual task and therefore to its biomechanical demands. Nevertheless, the mechanisms to achieve a precise control of the stabilization process in order to control the exerted force (Rancourt and Hogan, 2009) are not fully understood. Based on the fact that neuromuscular responses to mechanical instability are coupled to environmental dynamics, we can expect that control strategies such as muscle activation patterns are constrained by the particular mechanical condition of the task (Rancourt and Hogan, 2001a,b; Pinter et al., 2011). As pointed out by van Soest and van Galen (1995) the identification of such constraints is a prerequisite for proper understanding of the coordination of multi-joint movements. However, the aforementioned studies investigated either movements of the upper extremities or controlled, slow movements. The latter approach requires completely different use of stabilization-related and propulsive muscles as compared to fast movement velocities (Hagood et al., 1990). Here, we extend previous studies by investigating the effects of environmental stability changes during ballistic (high effort) multi-joint movements of the lower extremities (see also Wuebbenhorst and Zschorlich, 2012).

Consequently, this study is conceptually based on changes in the mechanical nature of the interaction against an external object. We assessed changes in the mechanics of muscular control by evaluation of intramuscular muscle activation characteristics. Having regard to previous observations of our laboratory, we hypothesized that (1) the force output would decrease with increasing external Degrees of Freedom (DoFs) and (2) that intramuscular coordination is adjusted to the type of mechanical interaction.

2. Materials and methods

2.1. Participants

The study involved 25 male volunteers (age: 24.8 ± 3.2 SD, height $180.2 \text{ cm} \pm 4.9$ SD, weight $82.7 \text{ kg} \pm 6.4$ SD). At the time of the study, none of the participants had ever experienced any injuries involving their lower extremities. The institutional ethics committee approved the experiments and all subjects gave written consent after being informed about the procedures of the experiment.

2.2. Apparatus

The subjects were asked to produce maximum force by performing an unilateral extension movement with their right leg in a sitting posture. The test apparatus used is shown in Fig. 1 and will be referred to as 'movement sled'. The sled moved on an inclined plane ($\alpha = 30^\circ$) with the movable parts having a weight of 18.2 kg. A counterweight of 37.5 kg (Pos. 2 in Fig. 1) was attached to the sled using steel cables and pulleys. The subjects were instructed to exert force against a footplate which varied in the

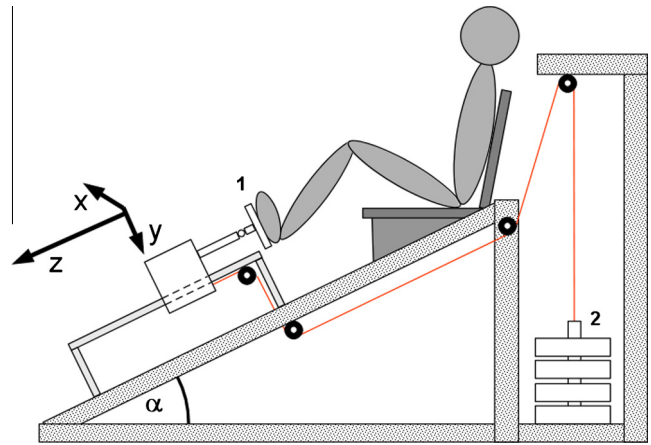


Fig. 1. The apparatus used in the present experiments (referred to as 'movement sled'). The sled moves on an inclined plane (angle $\alpha=30^\circ$). The footplate (pos. 1) was changed in order to create different experimental conditions. The counterweight (pos. 2) was set to 37.5 kg. Movement dimensions are indicated by arrows.

degree of instability (Pos. 1 in Fig. 1) to create different stabilizing conditions. In these experiments we used two dynamic conditions in which the sled was in an unhitched configuration. By using a rigid footplate (consisting of a block of wood), the subjects were exposed to 1 DoF. This condition allowed for movements along with the leg extension (z-direction in Fig. 1) and was considered to provide a stable interaction. In the second condition an unstable footplate was employed which was connected to the sled by a ball joint. Consequently, this condition caused 3 DoF, which comprised horizontal, rotational and translational movements that allowed for inversion/eversion and dorsal/plantarflexion of the ankle. The center of the ankle joint was aligned with the center of rotation of the ball joint of the device (footplate). This configuration was checked before every 3-DoF-trial and ensured the same point of contact between trials.

2.3. Protocol

The experiment was conducted in one session. Before performing the investigated movements, subjects completed a 5-min warm-up period on a treadmill (at 6 km/h) and 5–10 practice trials on the sled. All subjects were naïve to the test. In each condition (1 or 3 DoF) a block of five trials had to be completed, adding up to a total of 10 successful trials in the two conditions combined. The order of the tested conditions was randomized. This experimental approach allows to rule out effects of fatigue or the influence of testing order. The participants were asked to perform a ballistic, maximal leg extension (force production) movement against the sled. The criterion of a maximal effort in minimal time in each trial was eminent and each participant was constantly encouraged to do so. Participants were provided with constant verbal feedback about their performance and the force–time curve was checked in order to secure performance quality. For each trial the initial sitting position and the position of the limb (right leg) was checked so that each subject started with a knee and ankle angle of 90° . The correct limb arrangement on the sled was checked by the use of a goniometer. The seat was adjustable to the height of the subjects, which secured intersubject comparability. During data acquisition the subjects were to rest at least 30 s between trials to rule out potential fatiguing effects. Trials not matching our criteria of ballistic force production or trials with irregularities (e.g. slipping from the plate) were discarded from further analyses (judged by visual inspection of force–time curve).

2.4. Data recording

Surface electromyographic (SEMG) recordings were made from four muscles of the right leg: M. tibialis anterior (TA), M. peroneus longus (PL), M. vastus medialis (VM) and M. gastrocnemius medialis (GM). All surface EMGs were recorded using a bipolar setup (Wuebbenhorst and Zschorlich, 2011; Merletti & Hermens, 2004) with the reference electrode being placed in the middle of the recording electrodes for each muscle. Ag–AgCl electrodes (Hellige, GE Medical Systems, Germany; diameter: 6 mm) at an interelectrode distance of 3 cm were used for SEMG recordings. After standard skin preparation, the electrodes were attached to the skin in line with the direction of the muscle fibers. The electrode placement was consistent with the SENIAM recommendations (Hermens et al., 2000). Electromyographic signals were pre-amplified (1000 \times) and high-pass filtered at 10 Hz (Butterworth filter of second order, see Zschorlich, 1989).

For recording of force signals a three-dimensional Kistler force-transducer (Type 9602, Winterthur, Switzerland) with integrated charge amplifier was used (range 0–2000 N). The device was positioned directly behind the foot plate. The dimensions of force were defined as depicted in Fig. 1. All recordings were made at 2 kHz on a DAQ-Card (Type 6024E, National Instruments, Austin, Texas, USA). EMG signals were synchronized with the force data using a trigger of 200 N of the directional force–time curve (Fz) with a pre-trigger of 300 samples (150 ms).

2.5. Data analysis

In order to assess the activity generated by individual muscles we used an individually adjusted time window where we defined activity onset as the first positive rise of the rectified and smoothed (moving average smoothing filter with a time window of 25 ms) EMG signal from baseline (mean + 2 SD). This approach allowed for an individual muscle-based identification of activity. Movement end was defined as the first drop of the descending limb of the force–time curve under 10 N. The chosen time window had a variable length linked to individual movement performance and thus represents the relevant phase of stabilization (Wuebbenhorst and Zschorlich, 2011). Muscle activity was assessed by calculating the integral of the rectified and smoothed EMG–signal from movement onset to end. The force impulse exerted against the external object was calculated from the integral of the directional force–time curve (impulse of Fz). For that purpose movement onset was defined as the first rise of the ascending limb of the force time curve from baseline value (mean + 2 SD). Fig. 2 provides an overview about the performed analysis.

2.5.1. Muscle activity profiles

The effector system characteristics were assessed on the basis of the EMG–time curves. Three parameters were estimated: peak electrical activity (PEA), time to peak electrical activity (TTP) and slope of EMG. PEA represents the maximum of the rectified and smoothed EMG signal for each muscle using the same time window as used for calculation of muscle activity (see above). TTP typifies the time needed from the defined movement onset (see above) to reach peak electrical activity. Both parameters allow for assessment of amplitude (PEA) or temporal (TTP) scaling of muscular activity (see also Vorro et al., 1978; Vorro and Hobart, 1981a,b). In order to yield an adequate measure for characterization of total system behavior we used both parameters (PEA & TTP) to calculate the slope of muscle EMG (Vorro et al., 1978). This analysis was based on a linear approach for characterization of the steepness of the rise of muscular activity.

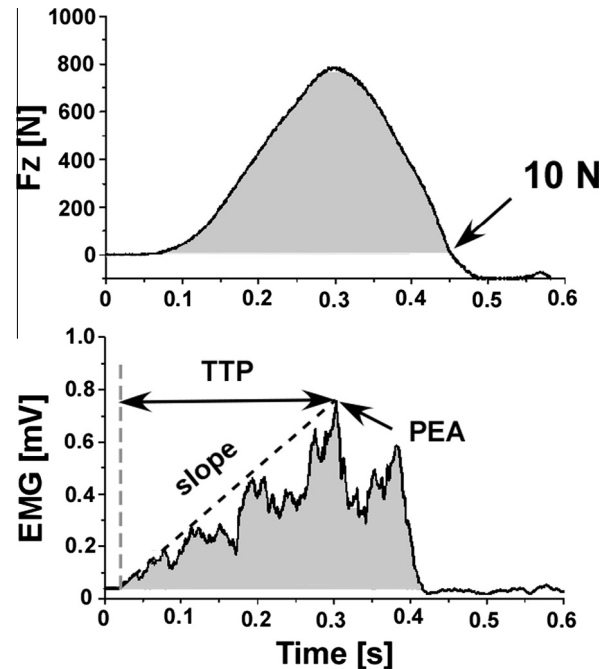


Fig. 2. The figure provides an overview about the analysis performed for assessment of magnitude of force exertion (upper panel) and EMG parameters (lower panel). The figure shows representative data of a single trial. The vertical dotted grey lines represent the defined onsets of movement. It has to be noted that the onsets were individually different for each muscle and for the force component Fz (see methods). Movement end was defined by a value of 10 N on the descending part of the force–time curve. The integrals of the force and EMG signals in the time window between movement onset and end were calculated (grey shaded area). The calculated myoelectric parameters included time to peak electrical activity (TTP), peak electrical activity (PEA) and slope.

2.6. Statistical analysis

In this study we assessed neuromuscular responses to changes in environmental dynamics by varying external degrees of freedom. We subjected the means of the force integrals and EMG–activity to a 1-way ANOVA, with DoF–condition (two levels: 1,3 DoF) as the independent factor. The individual dependent measures (PEA, TTP, slope) were analyzed using 2-way repeated measures ANOVA. The factors were DoF–condition (two levels: 1, 3), muscle type (four levels: TA, GM, PL, VM) and their interaction. For post hoc analysis we used Student’s *t*-test for pairwise comparisons between individual levels of the independent factors when warranted by a significant main effect. Post-hoc analysis was performed using Student’s *t*-Test where appropriate. Reported results are considered significant for *p*, at $\alpha = 0.05$.

3. Results

In the present study we investigated the modifications of myoelectric and dynamometric parameters in response to changes in environmental conditions. Our experimental design required subjects to perform a unilateral leg extension movement under two different mechanical conditions that varied in the degree of biomechanical demand.

3.1. basic performance measures

At first, the analysis of muscle activity showed a specific change in dependence of the DoF condition as reported in an earlier study of our laboratory (Wuebbenhorst and Zschorlich, 2011). This result is supported by significant differences between the DoF conditions

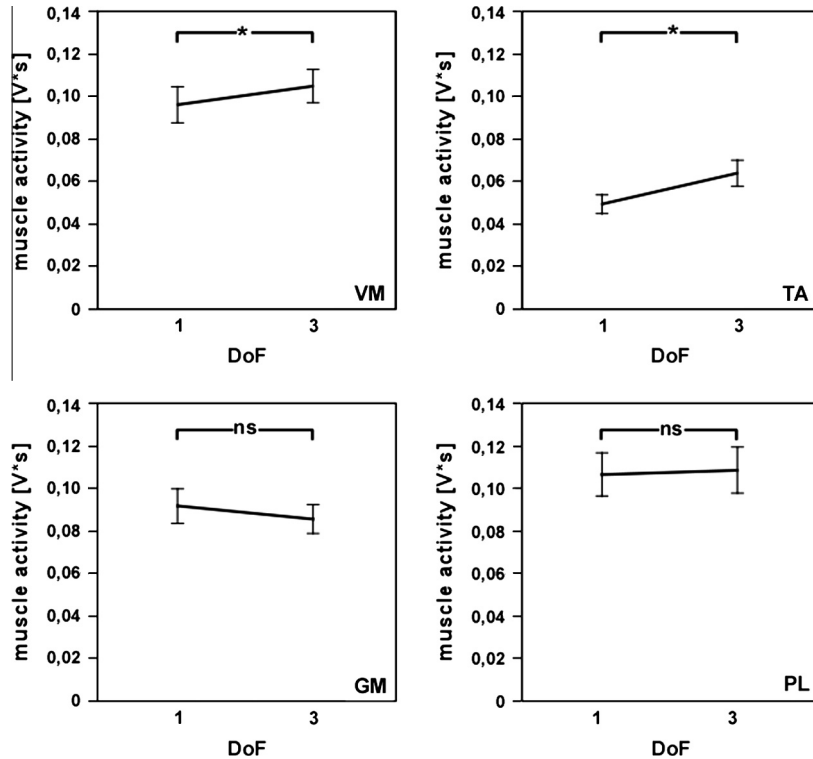


Fig. 3. Mean ± standard error for activities of muscles with motor function (VM, GM) and stabilizing function (PL, TA). The muscles showed an individual condition specific change in activity. (Significances $p < 0.05^{**}$, ns = non significant).

for the individual muscles (Fig. 3). For VM a significant elevation in activity emerged when increasing the external DoFs of the sled ($p < 0.05$). However, the GM and PL did not show altered pattern in comparison of the conditions (1 and 3 DoF) (GM: $p = 0.165$; PL: $p = 0.759$). In case of TA an enlarged electrical activity occurred between the 1 and 3 DoF ($p = 0.0042$).

These adjustments of activity levels by the individual muscles to environmental dynamics were accompanied by changes in force transmission to the surrounding (Fig. 4, Table 1). Consequently, the force impulse, calculated from the integral of the force time curve, revealed a decrease with the dynamic environmental conditions

Table 1

Muscle activities (upper four rows) in V*s and integral of directional force Fz (last row) in N*s. Individual values represent means ± standard error.

Conditions	1 DoF	3 DoF
GM	0.091 ± 0.008	0.085 ± 0.006
PL	0.106 ± 0.01	0.108 ± 0.011
TA	0.049 ± 0.005	0.064 ± 0.006
VM	0.096 ± 0.008	0.104 ± 0.009
Fz _{int}	151.8 ± 4.9	142 ± 4.96

becoming more unstable. This result was statistically confirmed by significant differences between the both conditions ($p = 0.004$).

3.2. muscle activity profiles

In order to assess the underlying neuromuscular basics of the reported changes in activity and force output we analyzed the activity profiles of the investigated muscles. Our analysis aimed on investigating intramuscular adjustments to task complexity.

With regard to the activity profiles of the individual muscles our data demonstrated significant interactions between DoF condition and muscle type for TTP [$F(3; 175) = 3.234, p < 0.0236$] but not for PEA [$F(3; 175) = 0.2314, p < 0.8744$]. The post hoc analysis revealed that the modulation of TTP varied across the muscles as a function of the particular DoF condition, while no changes at all were observed for PEA for all investigated muscles.

With respect to the temporal aspects of the intramuscular coordination process the results of the TTP measure of the TA muscle showed an antipodal gradient in comparison with the other investigated muscles (Fig. 5b). The control-scheme of TA showed a tendency for a temporal advanced peak of activity. However, the statistical analysis did not confirm the significance of the changes between 1 and 3 DoF (1–3 DoF: $p = 0.20$). In response to the

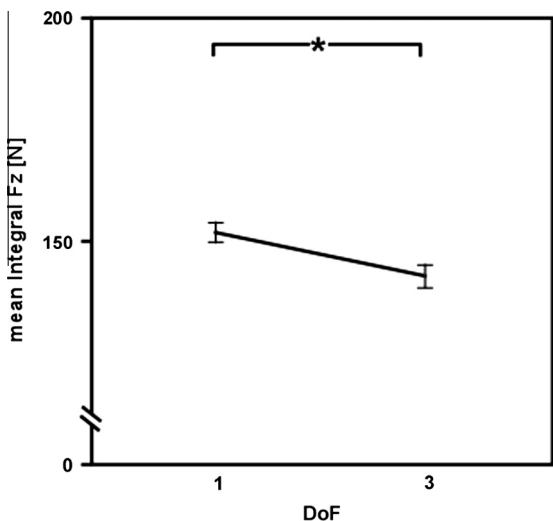


Fig. 4. Means ± standard error for integral of force in movement direction (Fz) sorted by particular DoF condition. A decrease of practicable force was noted when the external DoF was increased (Significances $p < 0.05^{**}$, ns = non significant).

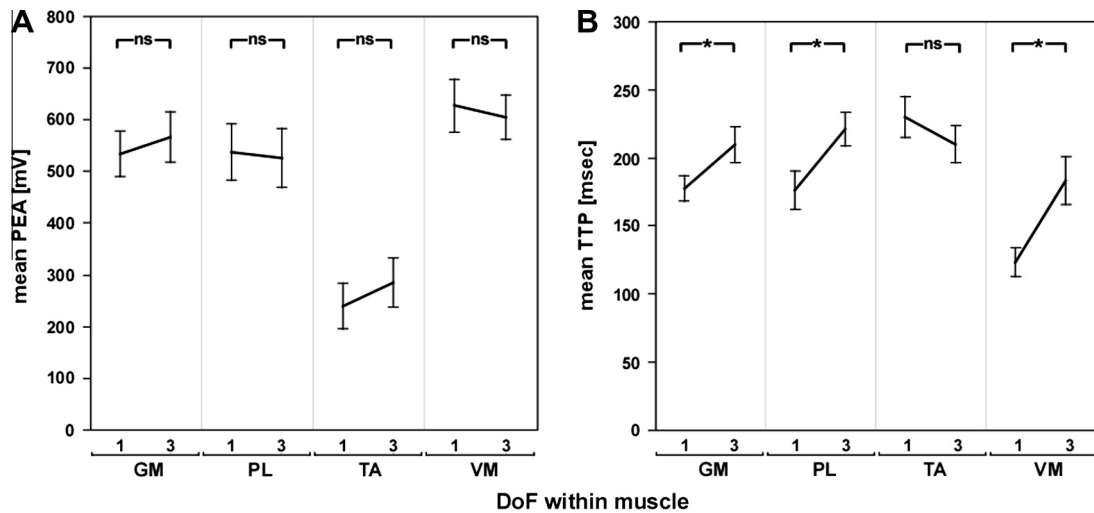


Fig. 5. (A) Mean \pm standard error for peak electrical activities (PEA) for all muscles influenced by the different DoF conditions. (Significances $p < 0.05$ **, ns = non significant). (B) Mean \pm standard error for time to peak (TTP) for all muscles. (Significances $p < 0.05$ **, ns = non significant).

Table 2

Means \pm standard errors for time to peak (TTP), peak electrical activity (PEA) and slope for all investigated muscles.

Muscle	DoF	Mean TTP (ms) \pm SE	Mean PEA (mV) \pm SE	Mean slope (mV/s) \pm SE
GM	1	177 \pm 9.51	532 \pm 43.7	3.3 \pm 0.32
GM	3	209 \pm 13.1	565 \pm 49.1	3.2 \pm 0.39
PL	1	175 \pm 14.2	536 \pm 54.6	3.6 \pm 0.42
PL	3	220 \pm 12.5	524 \pm 57.5	2.72 \pm 0.39
TA	1	229 \pm 15.2	238 \pm 44.5	1.07 \pm 0.18
TA	3	209 \pm 13.6	284 \pm 46.4	1.7 \pm 0.26
VM	1	122 \pm 10.6	626 \pm 50.9	6.66 \pm 0.74
VM	3	182 \pm 17.5	603 \pm 43.3	4.79 \pm 0.73

different experimental conditions the GM, VM and PL showed a significant delay when increasing the mechanical DoFs of the sled. For GM this delay is numerically reflected by an increase from 177 ms in the 1 DoF condition to a maximal value of 209 ms in the 3 DoF condition ($p = 0.0014$). Simultaneously, the time to reach peak activity of the VM was delayed by 60 ms in comparison of 1 (122 ms) and 3 DoF (182 ms) (1–3 DoF: 60 ms at $p = 0.0003$). In addition, the values for reaching peak activity of the peroneus longus muscle corresponded to 175 ms in the 1 DoF condition and 220 ms when the subjects were exposed to 3 DoF ($p < 0.0001$) (see Table 2).

Furthermore, we assessed the slope of EMG-signals of the investigated muscles. As shown in Fig. 6 the individual muscles showed specific changes in slope in dependence of the DoF-condition. Consequently, this measure revealed a significant interaction of DoF-condition and muscle type [$F(3; 175) = 4.306$, $p = 0.0059$]. On the basis of the individual muscles the VM revealed the most prominent changes in terms of a significant decline when increasing the external DoF ($p = 0.0004$). The biarticular GM did not show any alterations ($p = 0.2169$). In the case of TA and PL significant differences in slope emerged between 1 and 3 DoF, respectively (TA: $p = 0.022$; PL: $p = 0.0005$).

4. Discussion

This study investigated the effects of changing environmental dynamics on force transmission to the surrounding and neuromuscular control characteristics. Based on previous studies (Wuebbenhorst and Zschorlich, 2011; Holl and Zschorlich, 2011) we hypothesized that (1) the force transmission to the surrounding

would decrease with increasing external DoF and (2) the activity profiles would show adjustments depending on the type of environmental interaction. The first hypothesis was confirmed by the calculated force impulse relation between 1 and 3 DoF. Furthermore, the muscles showed a specific intramuscular tuning effect in dependence on the particular environmental condition, which thus supports the second hypothesis.

The finding of diminished force transmission is well in line with our previous observations that mastering an increased mobility of an external object is associated with an impaired ability to exert force effectively against that object. This effect is due to the necessity to stabilize the coupled mechanics of the object and the limb (Holl and Zschorlich, 2011). In extend of our previous finding of decreased peak forces with increased DoFs we analyzed the force impulse generated during task execution and our statistical analysis revealed a significant effect of DoF-condition. Our finding of decreased force output due to increased environmental dynamics is consistent with findings on wrist stabilization tasks involving different amounts of external DoF (Bober et al., 1982; Kornecki and Zschorlich, 1994; Kornecki et al., 2001).

Nevertheless, the change between 1 and 3 DoF was rather small (-6.8%) compared to our previous observations of maximal loss of

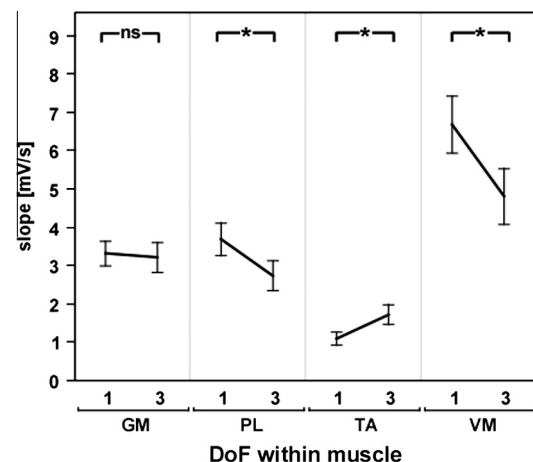


Fig. 6. Calculated values for slope (mean \pm standard error) for all investigated muscles separated by DoF condition, respectively. The VM muscle showed the most prominent changes expressed by decline of slope with increased DoF (Significances $p < 0.05$ **, ns = non significant).

peak force of -20.3% (Wuebbenhorst and Zschorlich, 2012). Thus, we reason that the neuromuscular system compensates for performance losses that are due to task requirements (as evidenced by reduced peak forces with increasing DoFs; Wuebbenhorst and Zschorlich, 2012) by adjusting the control strategy of the force exertion against the man-object system over time. This strategy might be used in order to maintain the ability to exert force as efficient as possible (or at least reduce the amount of force lost). This assumption implies that the control system provides a means for counteracting the destabilizing effect of force production (Rancourt and Hogan, 2009). Accordingly, the stiffness of the involved joints must be modulated with exerted force in a way that is constrained by task biomechanics (Rancourt and Hogan, 2001a,b).

Thus, as evidenced by the present study, the changes of force transmission to the surrounding were accompanied by altered control of effector system dynamics. In this context, the neuromuscular system adjusted the activity and control patterns of individual muscles in order to coordinate the interaction against the external environment.

This controlled interaction proved to be sensitive to the status of the external system on an intramuscular basis as evidenced from significant main effects for time to peak (TTP) and slope of muscle EMG. Regarding task execution, especially in the unstable 3 DoF condition the stability of movement was not secured by the environmental interaction, leaving the neuromuscular system to secure that stability. Thus, the necessity to provide a sufficient neuromuscular stiffness of the involved joints is eminent. The regulation of stiffness properties of joints plays an important role in order to compensate for instabilities, making the process of joint stabilization (at least) equally important as the force production process (Rancourt and Hogan, 2001a,b). Addressing the relation between stiffness properties and muscular activation, Hunter and Kearney (1982) pointed out, that muscle stiffness inherently scales with activation level. Additionally, this relation is also valid for high effort tasks reaching MVC-level (Weiss et al., 1988). Thus, adjustments of the motor apparatus to environmental instability in the present task are to be seen within the framework of stiffness regulation. However, in this sense stiffness regulation requires elevated co-contraction but, as shown before (Wuebbenhorst and Zschorlich, 2011, 2012), might be opposed by temporal coordination processes allowing a dosing of force output.

The changes of environmental dynamics provide the basis for the gain changes of the investigated muscles (Vorro and Hobart, 1981b). Thus, in order to maintain the ability to exert force against the sled, the muscles served for stabilization of the involved joints (knee and ankle) and (depending on muscle function) for propulsion. Nevertheless, as suggested in earlier studies from our laboratory (Wuebbenhorst and Zschorlich, 2011) and by Rancourt and Hogan (2009) the neuromuscular stiffness of joints is specifically tuned to offset the destabilizing effect of force production. As evidenced by the present study this tuning effect is linked to the coupled motion of the limb and the external object. Consequently, the particular role of individual muscles for orchestration of movement control varied across conditions.

The PL muscle did not change in electrical activity measures, but revealed a significant delay of peak activity (TTP) in the 3 DoF condition as compared to 1 DoF. Thus, we suggest a primarily temporal coordination strategy rather than modulation of activity level. Consequently, the slope of PL muscle EMG was significantly decreased between 1 and 3 DoF, suggesting a temporal gradation of activity by this muscle since the amount of activity of the PL did not change (as evidenced by PEA and iEMG values). Similarly, the GM revealed to PL comparable control characteristics, expressed for the most part in a delayed peak activity rather than scaling PEA or muscular activity. However, the slope of muscle

EMG did not reveal any changes between conditions. Consequently, these results underline the coordinative nature of the stabilization process also involving biarticular muscles like GM. Thus, these findings are well in line with earlier observations from our laboratory (Wuebbenhorst and Zschorlich, 2011) suggesting a dependency of muscular activation patterns of GM on primarily dynamic aspects of movement. Considering the biarticular nature of this muscle the temporal coordination is an expression of necessary means for controlling muscular force output, facing the fact that an inordinate activity might destabilize the movement (Hasan, 2005; Holl and Zschorlich, 2011). Additionally, as highlighted by Gomi and Osu (1998) the relative contribution of a biarticular muscle depends largely on the total behavior of all involved joints.

The VM muscle showed no changes in PEA, but revealed significant delayed TTP values when increasing external DoFs. Considering the fact that the muscle's activity (EMG impulse) increased significantly between conditions accompanied by a decrease in force output, the idea is supported that the VM not just contributes to leg extension (propulsion) but to the stabilization of the knee in this multi-joint task. The mechanisms used to control this functional coupling to actual task biomechanics are primarily temporal in nature. However, these adjustments resulted in a significant decline of slope when environmental dynamics became more demanding. This control strategy might reflect the necessity to control muscular force output of movement-producing muscles when the stability of movement is not provided by the environment. This assumption finds support from the idea that in order to create force under unstable conditions the involved joints must be stabilized first to keep the biokinematic chain stable (Wuebbenhorst and Zschorlich, 2011; Bober et al., 1982). Thus, the increase in total activity along with temporal adjustments point at a contribution of the VM muscle to both motor and stabilizing functions.

Finally, the motor characteristics of the TA muscle were dominated by increased muscular activity values accompanied by increased slope values in the 3 DoF condition. Thus, this muscle showed specific adjustments to task complexity expressed by a steeper rise of EMG activity in the unstable condition. These findings emphasize the role of the TA muscle for the stabilization process. The altered control and elevated maximal activity of the muscle in the unstable condition might allow for a rapid and sufficient compensation for potential disturbances induced by task execution. However, it has to be noted that this muscle was found to have a delayed onset of activity in an intermuscular comparison with the other investigated muscles (Wuebbenhorst and Zschorlich, 2011). Nevertheless, considering the fact that peak electrical activity was not different between conditions the current study demonstrates the importance of the TA muscle for contributing to overall coordination of the stabilization process.

Taken together, we observed that the PEA-values were not altered between stable and unstable experimental conditions for all investigated muscles directing the focus primarily on temporal coordination. Consequently, it is suggested that stabilization was not achieved by regulating joint stiffness through increased co-contraction but by adjusting temporal relations between muscles. However, it has to be noted that for VM and TA (both contributing to stabilization on the joints they are operating at) the iEMG measure were significantly different between conditions. Thus, regarding these two muscles the overall activity over the course of the movement was enhanced, but not PEA, confirming the idea of temporal contribution to regulation of stabilization.

4.1. Conclusion

This analysis contributes to the understanding of control modalities of multi-joint movements during interaction with different mechanical environments. The observed changes in

neuromuscular control characteristics reflect specific adjustments in controlling stiffness properties of the involved joints (see also Mueller and Blickhan, 2010) and to make the neuromuscular system less redundant by imposing constraints onto the system. This constraint-induced redundancy reduction is suggested to be implemented neurally as evidenced by task specific relative activation of the investigated muscles (Gielen et al., 1995). In addition to our previous finding of task specific intermuscular coordination strategies (Wuebbenhorst and Zschorlich, 2011) we found that intramuscular control is also linked to the type of mechanical interaction. Taken together these results imply that both inter- and intramuscular activation patterns must be controlled when interacting with environments of different stability increasing the number of variables to control during movement coordination. However, this strategy allows for flexibility in coordinating multi-joint movements (Gomi and Osu, 1998). This flexibility was mainly achieved by scaling temporal aspects (time to peak activity) of muscular activation, which is suggested to reflect a mechanism for dosing muscular force output for regulation of movement and stabilization. This assumption is supported by Barry et al. (2005) who attributed difficulties in motor performances exhibited by elderly to a decrease in flexibility of motor control strategies. Furthermore, adjusting muscular activity to the stabilization related demands of the task serves for modulation of joint stiffness. As shown in this study, the regulation processes used for solving this task are mainly time-dependent and may influence the magnitude of muscular activity created in response to different environmental dynamics.

Finally, this study is well in line with the idea of the impulse-timing hypothesis (Corcos et al., 1993) which states that changes in timing of muscle activity when quantity stays the same are responsible for changes in performance.

Conflict of interest

The authors have no conflict of interest to declare.

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