Program & Abstracts

7th Annual Meeting of the Danish Society of Biomechanics

Editor: Ernst Albin Hansen

SMI, Department of Health Science and Technology, Aalborg University

25-09-2015

WELCOME

Dear Colleagues

It is a pleasure for SMI and Department of Health Science and Technology, Aalborg University, to host the 7th Annual Meeting of the Danish Society of Biomechanics. The purpose of the annual meeting is to provide an opportunity for scientists, experts, and students to exchange and discuss research results concerning topics and approaches in the fields related to biomechanics within Denmark.

The annual meeting will consist of a Keynote Lecture, the Steno Lecture, as well as podium and poster presentations. Furthermore the annual General Assembly will be held.

We are looking forward to spending an interesting day with you!

On behalf of the organizers

Ernst Albin Hansen
DBS Board Member

MEETING INFORMATION

Keynote Speaker

We are very grateful that one of our conference sponsors, Qualisys, has made it possible to invite Dr Yuri Ivanenko as our Keynote speaker at the meeting.

Dr Yuri Ivanenko is Research Director, head of Gait Laboratory at the Fondazione Santa Lucia of Rome. He received his MS and PhD degrees in Biophysics from Moscow Physics and Technology Institute. He joined the Motor Control Laboratory at the Institute for Information Transmission Problems (Moscow), where he studied the mechanisms of muscle contraction and human posture control, was a lecturer in biophysics and biochemistry at the Ryazan Medical University (Russia) in 1986-91, did a post-doc at the Collège de France in Paris in 1995-98, and since then he has been working in the Department of Neuromotor Physiology at the Fondazione Santa Lucia. He is currently engaged in a large scale European project "Cogimon" investigating interactive human locomotion. His research interests include biomechanics and neurophysiology of human gait, development of locomotion, gait pathology, and motion perception. Yuri Ivanenko's articles have been published in various international journals including: J Neurosci, J Neurophysiol, Science, J Physiol, Nature Neurosci, J Exp Biol, Brain, J App Physiol, Exp Brain Res, PLoS One, Clin Biomech, Eur J Neurosci, Exerc Sport Sci Rev, Neuroscientist, Scand J Med Sci Sports, Curr Opin Neurobiol, etc.

Steno Speaker

The Steno lecture 2015 will be given by Professor Michael Voigt.

Professor Michael Voigt, who is with the Research Interest Group of Physical Activity and Human Performance, SMI, and Department of Health Science and Technology at Aalborg University, is this year's recipient of the Steno award. Professor Michael Voigt's main expertise lies in motor physiology and biomechanics, focusing on e.g. reflex function, movement related cortical potentials, 3D movement analysis, and muscle-tendon mechanics. He has published approximately 57 peer-reviewed articles and supervised 6 Ph.D. students and more than 15 master students in their master's projects.

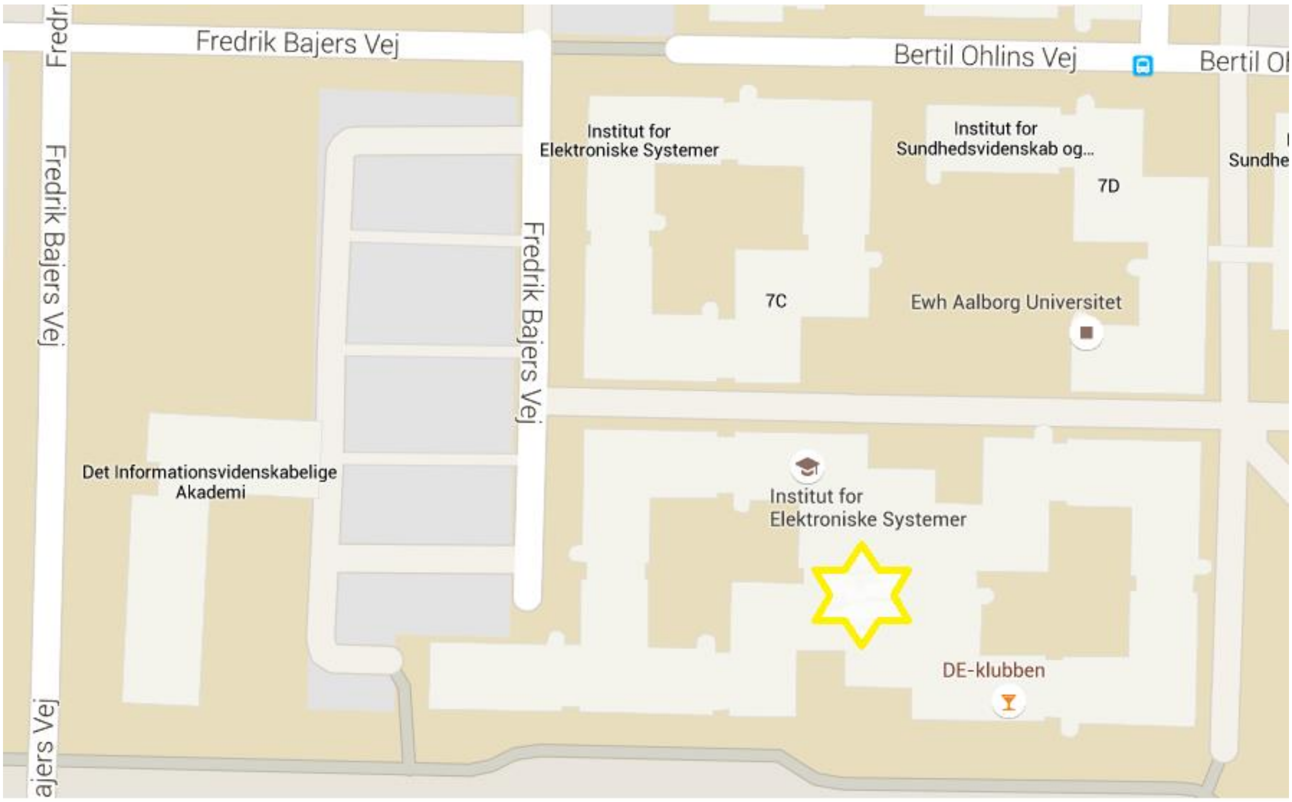
For more information see: http://personprofil.aau.dk/109545

Organizing Committee

Ernst Albin Hansen, Associate Professor, DBS Board Member
Uwe Kersting, Professor
Afshin Samani, Associate Professor
Christian Gammelgaard Olesen, Associate Professor
Mathias Vedsø Kristiansen, PhD student

Venue

Aalborg University
Fredrik Bajers Vej 7B, Auditorium B3-104 (indicated by a star on the map below)
DK-9220 Aalborg



All lectures, podium presentations, and the general assembly will take place in Auditorium B3-104. All other activities will take place in the area right outside the auditorium. Location for the activity of "Refreshments and networking" will be announced at the official closing of the meeting.

PROGRAM OVERVIEW

7th Annual Meeting of the Danish Society of Biomechanics	
9:15-9:45	**Registration, poster mounting, and coffee**
9:45-10:30	**Keynote lecture (Sponsored by Qualisys)**
10:30-11:15	**Steno lecture**
11:15-11:30	**Break**
11:30-12:30	**Podium presentations**
12.30-13:30	**Lunch**
13:00-13:25	**General assembly**
13:30-14:30	**Podium presentations**
14:30-15:40	**Posters and coffee (Sponsored by Qualisys)**
15:40-15:50	**DBS 2015 Student Award**
15:50-16:00	**Official closing of the meeting**
16:00-17:00	**Refreshments and networking**

PROGRAM DETAILED

9:15-9:45 **REGISTRATION, POSTER MOUNTING, AND COFFEE**

9:45-10:30 **KEYNOTE LECTURE (SPONSORED BY QUALISYS)**

Title: BIOMECHANICS AND NEURAL CONTROL OF HUMAN LOCOMOTION

By Yuri P Ivanenko

Chair: Ernst Albin Hansen

10:30-11:15 **STENO LECTURE**

Title: NEURO-MECHANICS OF STRETCH-SHORTENING MUSCLE ACTIONS IN LOCOMOTION MOVEMENTS

By Michael Voigt

Chair: Anders Holsgaard-Larsen

11:15-11:30 **BREAK**

11:30-12:30 **PODIUM PREESENTATIONS**

11:30-11:45 WALKING WITH AN EXOSKELETON: KINEMATIC DIFFERENCES COMPARED TO NORMAL WALK
Carsten Bach Baunsgaard, Peter Jarnved, and Tine Aljkær

11:45-12:00 PREDICTION OF MUSCLE-TENDON PARAMETERS BASED ON ISOKINETIC MEASUREMENTS
Frederik Heinen, Søren N. Sørensen, Mark King, Martin Lewis, Mark de Zee, and John Rasmussen

12:00-12:15 MUSCLE COORDINATION DURING BENCH PRESS IS CHANGED AFTER 5 WEEKS OF STRENGTH TRAINING – A RANDOMISED CONTROLLED STUDY
Mathias Kristiansen, Afshin Samani, Pascal Madeleine, and Ernst Albin Hansen

12:15-12:30 VARIABILITY PATTERN OF FORWARD BENDING OF THE TRUNK AMONG BLUE-COLLAR WORKERS
Morten Villumsen, Pascal Madeleine, Marie Birk Jørgensen, Andreas Holtermann, and Afshin Samani

Chair: Uwe Kersting

12:30-13:30 **LUNCH**

13:00-13:25 **GENERAL ASSEMBLY** (for DBS members)

13:30-14:30 **PODIUM PREESENTATIONS**

13:30-13:45 GAIT VARIABILITY AND MOTOR CONTROL IN PEOPLE WITH KNEE OSTEOARTHRITIS
Tine Alkjaer, Peter C. Raffalt, Helle Dalsgaard, Erik B. Simonsen, Nicolas C. Petersen, Henning Bliddal, and Marius Henriksen

13:45-14:00 AN ELECTROMYOGRAPHIC EVALUATION OF ELASTIC BAND EXERCISES TARGETING NECK AND SHOULDER PAIN AMONG HELM BEARIING MILITARY HELICOPTER CREW
Lars Askær Kristensen, Thomas Stig Grøndberg, Mike Murray, Gisela Sjøgaard, and Karen Søgaard

14:00-14:15 PREDICTION OF GROUND REACTION FORCES AND MOMENTS DURING RUNNING
Sebastian L. Skals, Moonki Jung, Michael Damsgaard, and Michael S. Andersen

14:15-14:30 BORN TO JUMP – MOTION PREDICTION USING FORWARD DYNAMICS
Søren Nørgaard Sørensen, and Frederik Heinen

Chair: Afshin Samani

14:30-15:40 **POSTERS AND COFFEE (SPONSORED BY QUALISYS)**

Board #1
THE PLACEBO EFFECT AND THE NUMBER OF ALTERNATIVE CHOICES CAN AFFECT MOVEMENT TIME DURING REACTIVE MOVEMENTS
Christian Sass Axelsen, Kasper Simonsen, Martin Pedersen, Simon Husted, and Michael Voigt

Board #2
EMG EVALUATION OF THREE ELASTIC BAND EXERCISES FOR BEDBOUND PATIENTS
Tina Dalager, Stine Stensgaard, Karen Søgaard, and Gisela Sjøgaard

Board #3
THE ASSOCIATION BETWEEN ECCENTRIC HIP ABDUCTOR STRENGTH AND KINEMATIC MARKERS ASSOCIATED WITH EITHER ILIOTIBIAL BAND SYNDROME OR PATELLO-FEMORAL PAIN: A CROSS SECTIONAL STUDY ON RUNNING
René Korsgaard, Sten Rasmussen, Rasmus O. Nielsen, Uffe Laessoe, and Michael Voigt

Board #4
CLASSIFICATION OF FOOT STRIKE PATTERN WITH INCREASING RUNNING SPEED
Cilie Lindner, Maria T. Hansen, Martin S. Manstrup, and Michael Voigt

Board #5
DESIGN OF A SUBJECT-SPECIFIC AMERICA'S CUP GRINDING HANDLE: FROM 3D-SCANNING TO 3D-PRINTING TECHNIQUES – A TECHNICAL NOTE
Morten Bilde Simonsen, Anders Rosendal Jensen, Miguel Nobre Castro, and Christian Gammelgard Olesen

Board #6
KINEMATIC AND KINETIC EFFECTS OF VARIABLE RESISTANCE ON THE CONVENTIONAL DEADLIFT
Markus Sloth, Gorm Rasmussen, Afshin Samani, Michael Mortensen, and Kasper Nielsen

Board #7
DEVELOPMENT OF A WORKFLOW FOR WEAR AND FUNCTIONAL SIMULATION OF TOTAL KNEE ARTHROPLASTY
Jonas Stensgaard Stoltze, Mark Taylor, John Rasmussen, and Michael Skipper Andersen

Board #8
DOES KINESIOTAPE FACILITATE OR INHIBIT THE ACTIVATION OF THE TRAPEZIUS MUSCLE?
Rasmus Tran, Lise B. Johansen, Mia Laursen, Michael K. Festersen, and Pascal Madeleine

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15:40-15:50 **DBS 2015 STUDENT AWARD**

Chair: Uwe Kersting

15:50-16:00 **OFFICIAL CLOSING OF THE MEETING** (Ernst Albin Hansen)

16:00-17:00 **REFRESHMENTS AND NETWORKING**

ABSTRACTS

GAIT VARIABILITY AND MOTOR CONTROL IN PEOPLE WITH KNEE OSTEOARTHRITIS

Tine Alkjaer[1*], Peter C. Raffalt[1], Helle Dalsgaard[1], Erik B. Simonsen[1],

Nicolas C. Petersen[1,2], Henning Bliddal[3], Marius Henriksen[3]

[1]Department of Neuroscience and Pharmacology, University of Copenhagen, [2]Department of Nutrition and Exercise, University of Copenhagen, [3]Clinical Motor Function Laboratory, The Parker Institute, Department of Rheumatology, Copenhagen University Hospitals Bispebjerg and Frederiksberg, Nordre Fasanvej 57, 2000 Frederiksberg, Denmark

INTRODUCTION

Knee osteoarthritis (OA) is a common disease that impairs walking ability and function. We compared the temporal gait variability and motor control in people with knee OA with healthy controls. The purpose was to test the hypothesis that the temporal gait variability would reflect a more stereotypic pattern in people with knee OA compared with healthy age-matched subjects.

METHODS

We included patients with a clinical diagnosis of knee OA (radiographically verified) from the OA outpatient clinic at Copenhagen University Hospital Bispebjerg and Frederiksberg. Healthy age-matched volunteers were recruited among colleagues and relatives of employees at the University of Copenhagen.

The experiment consisted of two parts: 1) Assessment of soleus (SO) H-reflex modulation during treadmill walking at 3.5 km/h and 2) assessment of the gait pattern variability and muscle activities during continuously walking for 6 min on the treadmill at 3.5 km/h.

The gait variability was assessed from goniometry of the ankle and knee joints and gait events recorded by from foot switches placed under the heel and toe. The temporal structure of the ankle and knee joint kinematics was quantified by the largest Lyapunov exponent and the stride time fluctuations were quantified by sample entropy and detrended fluctuation analysis. In addition, co-contraction between the ankle and knee muscles pairs was calculated.

RESULTS AND DISCUSSION

Fifteen OA patients (females) met the inclusion criteria and were invited to participate of which 11 accepted the invitation. Four were unable to participate for practical reasons. One OA participant aborted the H-reflex measurements due to discomfort during the stimulations, but completed the gait variability assessments and EMG recordings. Eleven healthy age-matched female subjects were recruited as controls.

The results showed no statistically significant mean group differences in any of the gait variability measures or muscle co-activation levels. The SO H-reflex amplitude was significantly higher in the knee OA group around heel strike when compared with the controls (Fig. 1). The mean group difference in the H-reflex in the initial part of the stance phase (control–knee OA) was -6.6 %Mmax (95% CI: -10.4 to -2.7, p=0.041). The present OA group reported relatively small impact of their disease.

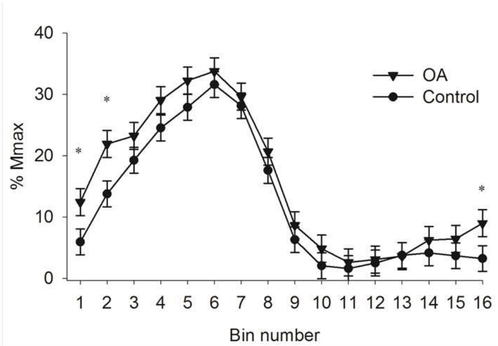

Fig. 1 The SO H-reflex modulation over the gait cycle averaged in 16 bins (stance phase bin 1-10 and swing phase bin 11-16) of the OA (filled triangles) and control groups (filled circles). Error bars are ±SE. Asterisks indicate statistical significant differences between the groups at p<0.05.

CONCLUSIONS

These results suggest that the OA group in general sustained a normal gait pattern with natural variability but with suggestions of facilitated SO H-reflex in the swing to stance phase transition. We speculate that the difference in SO H-reflex modulation reflects that the OA group increased the excitability of the soleus stretch reflex as a preparatory mechanism to avoid sudden collapse of the knee joint which is not uncommon in knee OA.

ACKNOWLEDGEMENTS

This study was supported financially by The Danish Rheumatism Association, grant no. R71-A1020.

The placebo effect and the number of alternative choices can effect movement time during reactive movements

Christian Sass Axelsen, Kasper Simonsen, Martin Pedersen, Simon Husted, Michael Voigt
Department of Health Science and Technology, SMI®, Aalborg University

INTRODUCTION

The effect of caffeine on the central nervous system is linked to the fact that caffeine blocks the adenosine receptors placed in the synapses. Adenosine is known to slow down the neuronal activity (1). According to this information caffeine can improve reaction time and create a better environment in the synapse knowing that adenosine will not be able to slow down the neuronal activity in the same way. This improved environment could also be affecting other factors as movement time.1) caffeine intake of 400 mg and placebo will improve reaction time. 2) caffeine intake of 400 mg or placebo will not improve movement time. 3) the movement time will not be improved by numbers of alternative choices.

METHODS

13 physically active men (mean age 24±1.04 and BMI<30). The participants were required to visit the laboratory three times for baseline, placebo and caffeine reaction time testing respectively within a test period of one week to complete three tests at each visit. The three tests included: a one-choice, a two-choice and a four-choice reaction time test. The order of administration of placebo and caffeine as well as the order of the two-choice and four-choice tests were randomized. In the one-choice test the target extremity for the choice was the right leg, in the two-choice in test right and left leg were target extremities and in the four-choice test both arms and both legs were targets of choice. The experimental set-up consisted of an aluminum frame with 4 photocells mounted (two for arm reactions and two for leg reactions).The frame was placed just in front of two force platforms on which the participants were placed. The reaction time was measured as the time from a 'go signal' to the first reaction detected with the force platforms. The movement time was measured as the time of the first reaction detected with the force platforms and the time a given photocell closed.

In all tests the time from a ´ready-signal' to the 'go signal' was randomized. The 'ready-signal' and 'go-signal' were presented by LED's mounted on a small box placed on the aluminum frame just in front of the participant at the level of the eyes.

These results were found using a two way repeated measure ANOVA.

RESULTS AND DISCUSSION

The results in figure 2 shows that movement time was decreased by the intake of placebo and caffeine. Although no significantly difference between placebo and caffeine, could indicate that only a placebo effect was determinated. According to the alternative choices the movement time was not affected in every manipulation the subjects were exposed to. Only when the difference between the numbers of alternative choices of movement increases from one choice to four different choices, the movement time decreases in both the baseline, placebo and caffeine test.

CONCLUSIONS

Caffeine administration had no significant effect on either movement time or reaction time although placebo resulted in an improvement in the movement time. The number of alternative choices of movement was also affecting the movement time.

REFERENCES

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2. Kroemer, K. H. E., H. B Kroemer, and K. E. Kroemer-Elbert. Ergonomics: how to design for ease and efficiency. 2nd ed. Prentice Hall international series in industrial and systems engineering. Upper Saddle River, NJ: Prentice hall. 137-139, 2001

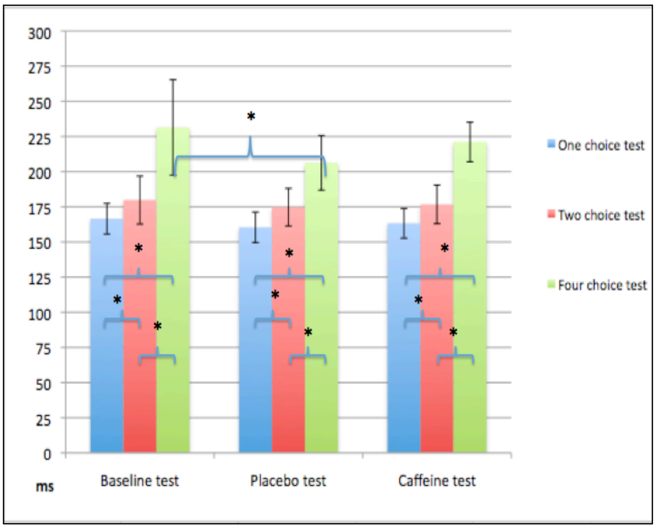

Figure 1: The mean reaction time (*: p<= 0.05)

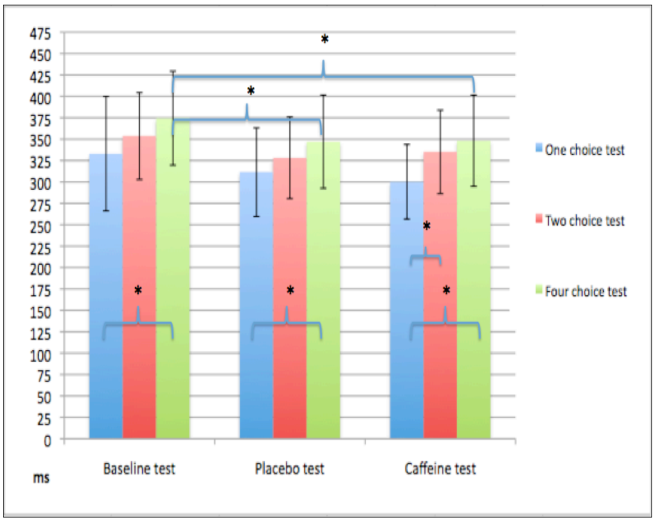

Figure 2: The mean movement time (*: p<= 0.05)

WALKING WITH AN EXOSKELETON: KINEMATIC DIFFERENCES COMPARED TO NORMAL WALK

Carsten Bach Baunsgaard, PhD student[1,2*], Peter Jarnved [3], Tine Aljkær[1]

[1]Department of Neuroscience and Pharmacology, Panum Institute, University of Copenhagen, , [2]Department of Spinal Cord Injuries, Rigshospitalet, [3]Danmarks Tekniske Universitet DTU; e-mail:cbbaunsgaard@sund.ku.dk

INTRODUCTION

In this study we investigate a robotic othosis called an exoskeleton, which is a wearable robotic orthosis for use in neurological rehabilitation. These robotoc devices have moved from test laboritories to the rehabilitation clinics and are now being included in rehabilitation practice worldwide. It is predicted that in the near future they will be far more widespread also in private use for mobility devices. This widespread use makes it important to characterise how ambulation with these devices divert from normal gait, in order to predict the potentials and challenges this new paradigm shift in rehabilitation.

In this study we test how the kinematics of gait of healthy individuals change from walking freely without assistive devices, to walk controlled by the exoskeleton.

METHODS

All participants were healthy, without previous neurological disorders or musculoskeletal problems.

Ten participants were chosen as sample of convenience and were their own controls. We used the exoskeleton Ekso® GT, from Ekso Bionics. Goniometer data from hip, knee and ankle on the dominant side are recorded after calibration of the goniometer. Ankle joint in the Ekso is without motor actuators and built with low range of motion, so this was chosen not to include in the study.

Participants walked under three different conditions:

1) Without Ekso, self-selected pace and stride length.
2) Without Ekso, controlled stride length by markers on the floor and pace by metronome. Stride length and pace was set to be the same as the Ekso settings.
3) With the Ekso, same stride length and pace as condition two. Before recording participants had a habituation period of 15 min of walking with the Ekso before recording.

Goniometer data from 10 steps is averaged from each condition.

RESULTS AND DISCUSSION

Participant's characteristics (median, IQR): Age, gender, height, weight.

Gait speed in condition two and three was 1.97 km/h (stride 70 cm, pace 47 bpm).

Relative ankles of the hip and knee were significantly larger in hip and knee during walking with the Ekso than walking freely. Gait speed was lower than normal preferred speed.

	Normal	Ekso	t-test
ROM hip(°)	18,1	39,3	$p<0.05$
ROM knee(°)	41,7	57,2	$p<0.05$

This is preliminary results. Data will be analyzed further with more advanced kinematic analysis, including data from ankle dorsi- and plantar-flexion. Electromyographic data from four muscles in the lower extremity will be added to the analysis to characterize the muscle involvement and timing of these.

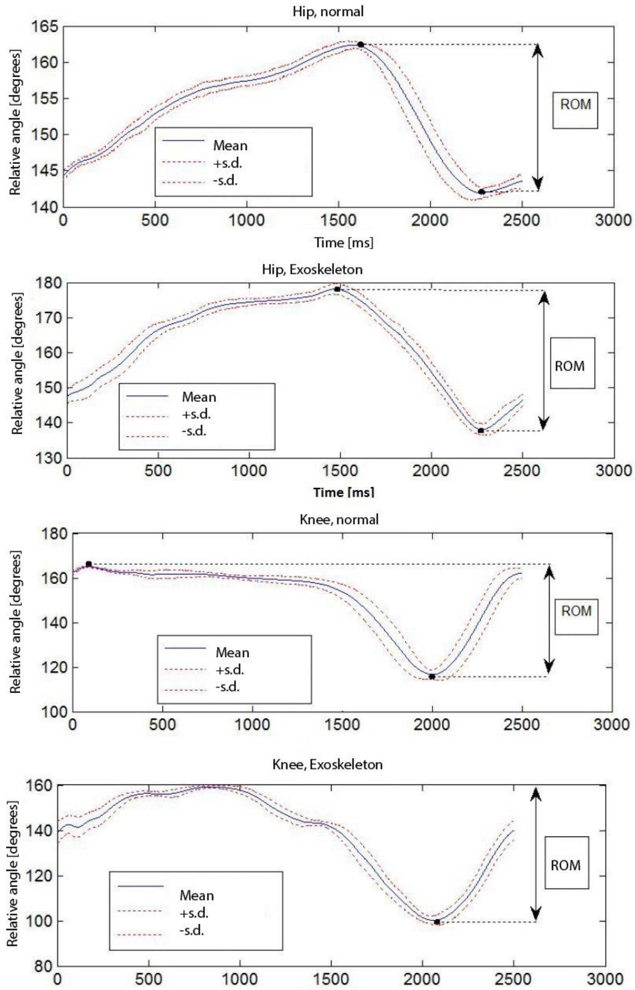

Fig. 1 changes in ROM during gait cycle, with and without the Ekso.

CONCLUSIONS

Gait with an exoskeleton orthosis changes the gait kinematics significantly

ACKNOWLEDGEMENTS

We would like to thank the participants of the study.

EMG EVALUATION OF THREE ELASTIC BAND EXERCISES FOR BEDBOUND PATIENTS

Tina Dalager[1*], Stine Stensgaard[1], Karen Søgaard[1] & Gisela Sjøgaard[1]

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INTRODUCTION

Elderly medical patients who are being hospitalised in Denmark spend up to 17 hours a day in bed [1]. Besides the risks for decreased balance, lung complications and bedsore, even few days of immobility induces a loss of muscle mass and muscle strength [2,3]. Strength training is shown to counteract these effects [4,5]. However, being hospitalised and newly operated limits the choice of suitable exercises and training equipment.

Therefore, the aim of this study was to evaluate muscle activity during three elastic band exercises carried out in the hospital bed.

METHODS

Six healthy females (mean age 45, range 27-56 yrs., mean BMI 23, range 20-27) volunteered. Three elastic band strength exercises (shoulder extension, elbow extension and hip extension) using Thera-Band™, were evaluated. All exercises were performed while lying in bed with the elastic band fixed in the ceiling. Electromyography (EMG) was measured on 8 relevant muscles (5 upper body, 3 lower body muscles) with wireless MYON 320 (Myon AG Switzerland) and shown as mean EMG activity in percentage of Maximum Voluntary Electrical activation (MVE) for each muscle. Individual 8 and 15 Repetition Maximum (RM) was pre-detrmined by a combination of elastic band resistance and length of the elastic band. Rate of perceived exertion (RPE) was measured using the Borg 10-point RPE scale.

RESULTS AND DISCUSSION

Figure 1 shows for the three different exercises the mean EMG in percentage of MVE for 15RM and 8RM. Shoulder extension and elbow extension primarily activated m. triceps brachii, and hip extension primarily activated m. biceps femoris. No significant difference between 15RM and 8RM was found for any of the three exercises.

RPE for the three exercises was as follows with no significant difference between 15RM and 8RM:

	Shoulder extension	Elbow extension	Hip extension
15RM	8.3 ± 1.1	8.7 ± 1.0	8.3 ± 1.0
8RM	9.0 ± 0.8	8.7 ± 1.1	8.3 ± 1.3

Table 1: RPE values for the three exercises

This is the first study using a fix-point in the ceiling for in-bed elastic exercises. This, in combination with different levels of elastic band resistance and lengths makes an individual's adaptation to strength training feasible, even for bedbound patients. Furthermore the EMG and RPE measurements show that the three chosen exercises target relevant muscle groups. Surprisingly, we found no statistical significant differences between 15RM and 8RM in muscle activation. This might be due to the difficulty of identifying specific RM for elastic exercises.

CONCLUSIONS

The study advocates using intensive strength training during a hospital stay. It is of great importance to look for new ways to facilitate a more optimal recovery period among elderly postoperative patients.

REFERENCES

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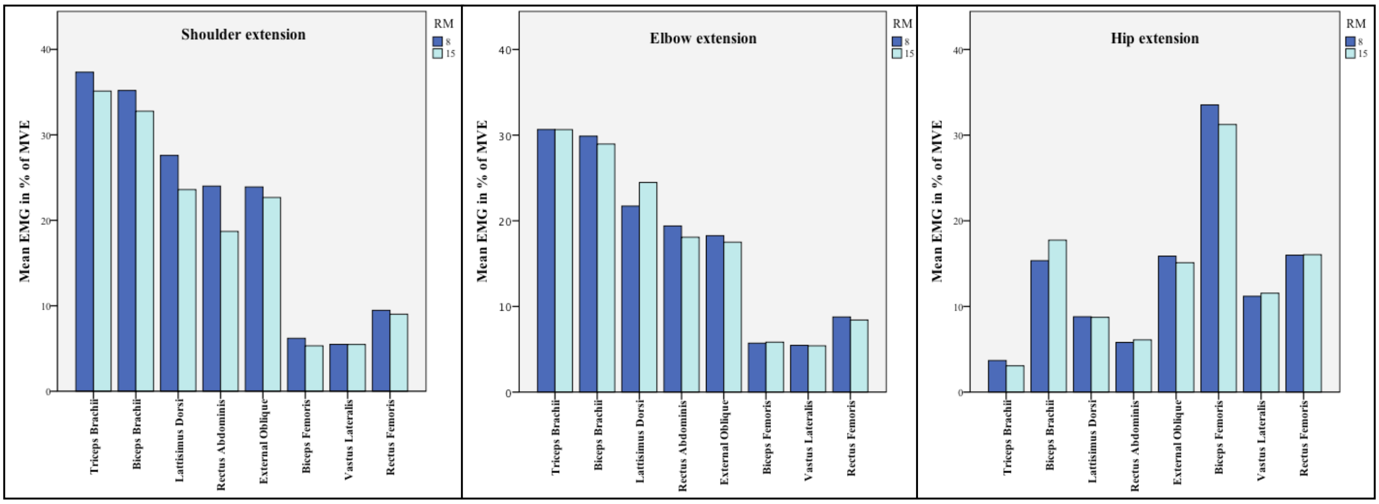
5. Suetta C et al., *Scand J Med Sci Sports* **17(5)**: 464-472, 2007

Figure 1: Mean EMG in percentage of MVE (SD)



PREDICTION OF MUSCLE-TENDON PARAMETERS BASED ON ISOKINETIC MEASUREMENTS

Frederik Heinen[1,2,*], Søren N. Sørensen[2], Mark King[3], Martin Lewis[4], Mark de Zee[1] & John Rasmussen[2]

[1]Department of Health Science and Technology, Aalborg University, [2]Department of Mechanical and Manufacturing Engineering, Aalborg University, [3]School of Sport, Exercise and Health Sciences, Loughborough University, [4]School of Science & Technology, Nottingham University; e-mail:fh@hst.aau.dk, *Ph.D. student

INTRODUCTION

Musculoskeletal models are often based on scalable generic models that implement phenomenological Hill-type muscle models [1]. The force-producing capability of the Hill model is especially sensitive to certain parameters [2], which do not scale linearly with external body dimensions.

The aim of this study was to determine and analyze the muscle-tendon parameters based on experimentally obtained isokinetic measurements.

METHODS

One male long distance runner (height 1.85 m and mass 66.5 kg) was included in this study, which was carried out in accordance with the Loughborough University Ethical Advisory Committee guidelines. A series of isometric and isokinetic measurements were performed on the dominant leg for the ankle, knee and hip flexors/extensors using a Contrex multi-joint isovelocity dynamometer (CMV AG, Switzerland). Seven evenly spaced isometric measurements were performed for each joint motion throughout the subject's joint range-of-motion (ROM). Six (ankle and hip) and eight (knee) isovelocity measurements from 50°/s and increasing in increments of 50°/s were performed in both concentric and eccentric contraction. The measurements were corrected for gravity and passive torques components throughout the ROM using built-in software. A lower extremity model was adopted based upon the TLEMsafe 2.0 model [3] using the commercially available software, AnyBody Modeling System (AnyBody Technology A/S, Denmark). The model was scaled based on anthropometric measurements. A series of models were created mimicking the experimental conditions to evaluate the net joint strength at the different angles and velocities. The muscle recruitment was based on the min-max criterion to ensure maximal utilization of maximum strength from the model. Optimization procedures were adopted in Python using the SNOPT optimizer with standard settings in the open-source software PyOpt. The optimization procedure was based on the method presented by Garner and Pandy [4] and minimized the difference between the experimental and simulated isometric joint strengths:

$$minimize\left(J\left(\tilde{L}^m_{max}, \tilde{L}^m_{min}, F_{local}\right) = \sqrt{\sum_{i=1}^{n}(\tilde{T}_i^{exp} - \tilde{T}_i^{mod})^2 + P}\right)$$

Where n was the number of considered muscles (flexors/extensors), $\widetilde{\mathrm{T}}_{\mathrm{i}}^{\mathrm{exp}}$ and $\widetilde{\mathrm{T}}_{\mathrm{i}}^{\mathrm{mod}}$ are, respectively, the experimental and model-predicted net joint strengths for the *i*'th joint position/velocity normalized to the maximal experimental joint torque. Two design variables were used for each muscle: two normalized fiber lengths, one shorter ($\tilde{L}^m_{min}$) and one longer ($\tilde{L}^m_{max}$) than optimal fiber length. A local strength factor (F_{local}) was assigned for the whole muscle group altering the peak isometric force. The bi-articular muscles were assigned with the average of the involved local strength factors. The objective function was augmented by a penalty function (P) that adds a large, positive value in case the optimizer tries to assign a muscle with a non-physiologic short tendon or fiber lengths.

RESULTS AND DISCUSSION

The isometric optimization procedure resulted in a relatively good fit between the experimental and modeled joint strengths for the ankle plantar flexors (Figure 1). The root mean squared difference (RMS) between the experimental and respectively the reference model (RMS 58.1%) and optimized model (RMS 6.6%) decreased.

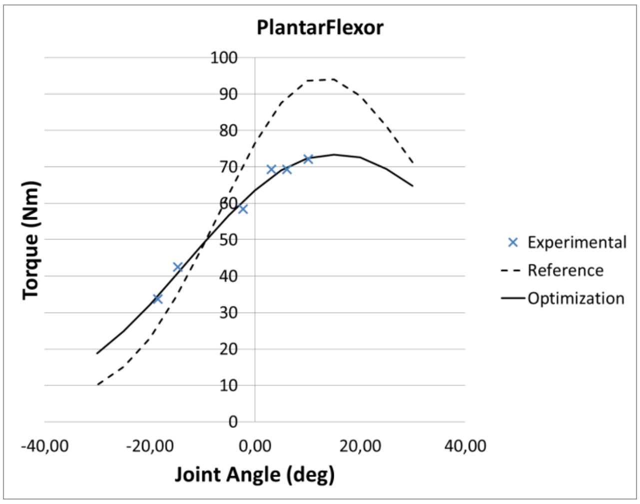


Fig. 1 A comparison of the experimental, reference model and optimized model torque-angle curves for the plantar flexors.

CONCLUSIONS

Good agreements were observed between the experimental torques and optimized modeled torques. Moreover, it is important to note that scaling based on individual measurements leads to a big improvement relative to the general scaling.

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BIOMECHANICS AND NEURAL CONTROL OF HUMAN LOCOMOTION

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The idea that the CNS may control complex interactions by modular decomposition has received considerable attention. We explored this idea for human locomotion by examining the functional structural organization of the motor output. The results suggest that the brain employs a modular strategy, coordinating whole limbs rather than their individual components. Furthermore, the coordination of locomotion with voluntary tasks may be accomplished through a superposition of motor programs or activation timings that are separately associated with each task. As a consequence, the selection of muscle synergies appears to be downstream from the processes that generate activation timings. Each human lower limb contains over 50 muscles that are coordinated during locomotion and many of them (for instance, intrinsic foot muscles) indicate a high-level of task-dependent specificity in their function. Motor programs in a variety of locomotion conditions may be considered as a characteristic timing of muscle activations linked to specific kinematic events, and the spatiotemporal maps of spinal cord motoneuron activation also show discrete periods of activity. Their timing and duration become tuned during development to our unique heel-strike gait. Biomechanical correlates of each activation pattern have been described, leading to the hypothesis that the co-ordination of limb and body segments arises from the coupling of neural oscillators between each other and with limb mechanical oscillators. Ongoing debate and the current discussion of the critical aspects and organization of basic activation patterns will be considered. Recent findings from other perspectives on modularity, namely the developmental and evolutionary perspective, will also be presented.

THE ASSOCIATION BETWEEN ECCENTRIC HIP ABDUCTOR STRENGTH AND KINEMATIC MARKERS ASSOCIATED WITH EITHER ILIOTIBIAL BAND SYNDROME OR PATELLO-FEMORAL PAIN: A CROSS SECTIONAL STUDY ON RUNNING

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INTRODUCTION

Several kinematic angles have been associated with iliotibial band syndrome (ITB) or patello-femoral pain (PFP). Knee internal rotation, hip internal rotation and hip adduction are proposed to be associated with both ITB and PFP, while ITB has also been associated with knee adduction and PFP has conversely been associated with knee abduction during locomotion [1, 2]. Weak eccentric hip abductor strength (EHAS) has been associated with increased knee adduction due to contralateral pelvic drop; thereby increasing the horizontal distance the center of mass (CoM) has to the knee joint center [3]. Some runners compensate for this by moving the CoM closer to the stance limb. Moving the CoM excessively towards the stance limb might, however, change the knee to abduct instead [3]. The associations between EHAS and the kinematic markers associated with PFP (PFPM group) or ITB (ITBM group) may help clinicians select whether to strengthen the hip abductors in injured runners based on the running kinematics. Therefore, the aim of this study was to determine the association between EHAS and the kinematic markers associated with either ITB or PFP.

METHODS

One hundred male recreational runners were recruited for this study. The runners ran at least two times a week, have at least two years of running experience, and have been without any pain for the last three months. Their mean (± standard deviation) age was 37 ± 11 years, with a mean height and weight of 182 ± 10 cm and 79 ± 10 kg, respectively. EHAS and running kinematics were measured on both legs, with each leg considered as independent. EHAS was obtained with an isokinetic Biodex dynamometer and running kinematics was collected in 3D with the Codamotion active marker system. EHAS data was lost for seven participants. A linear regression between EHAS and these two groups of runners was performed. The variables were checked visually for a linear relationship and outliers using a scatterplot of the dependent variable and the explanatory variables. The homoscedasticity and normal distribution was checked using a p-p plot. Due to concerns about the data being right-skewness, a sensitivity analysis was performed using robust variance estimation and a bootstrap with 1000 replications to confirm the confidence interval range.

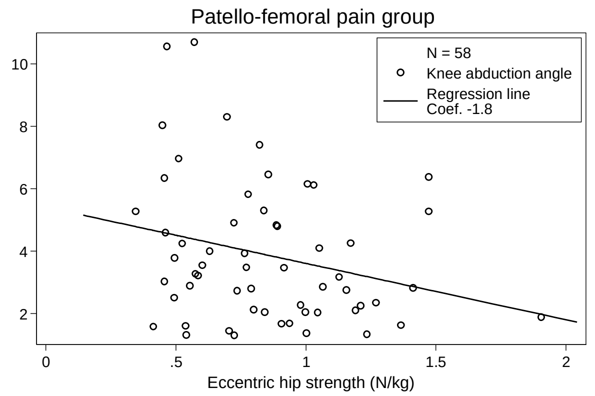

Fig. 1 Regression analysis on the association between EHAS and knee abduction for the PFPM group only. The PFPM group comprised those runners displaying knee abduction during the stance phase of running**.** 58 legs were displayed knee abduction and 128 legs displayed knee adduction (not shown).

RESULTS AND DISCUSSION

Linear regression on the association between EHAS and running kinematics of either the ITBM group (128 legs adducting the knee) or the PFPM group (58 legs abducting the knee) was performed. Only the point estimate of the association between EHAS and knee abduction (-1.80; 95% CI -3.59; -0.009) in the PFPM group was significant (Fig. 1). The present study has demonstrated an association between EHAS and knee abduction in runners with kinematic markers associated with PFP. Therefore, increasing the strength of the hip abductors might be beneficial for runners with PFP and a visible knee abduction angle during stance.

CONCLUSIONS

Our results demonstrate an association between EHAS and knee abduction angle in runners with kinematic markers associated with PFP. However, the causality between EHAS and PFP still needs to be fully clarified.

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An electromyographic evaluation of elastic band exercises targeting neck and shoulder pain among helm bearing military helicopter crew.

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INTRODUCTION

Flight related neck and shoulder pain is a frequent problem in helicopter pilots and crew [1]. Pain causes personnel suffering, reduces operational capabilities and incurs high financial cost due to the loss of manpower. Evidence suggests that the occupational loading such as posture adopted during flight and increased weight added to the mass of the head due to the helmet and night vision equipment contribute to the development of neck and shoulder pain. Strength training has among other occupational groups been found to reduce musculoskeletal pain [2]. A 20-week exercise program for the neck and shoulder muscles using elastic bands has been applied for helicopter pilots and crew in the Royal Danish Air Force to prevent and reduce pain. The exercise program had an initial loading of 20RM and was increased progressively towards 12RM in the final weeks. A muscle activity >60% MVE is generally considered necessary for gaining strength, but with elastic bands it is hard to control exercise intensity [3]. The purpose of the present study was to conduct an electromyographic evaluation of the neuromuscular activity in trapezius descendens, sternocleidomastoideus and the upper neck extensors during an elastic band exercise program targeting neck and shoulder pain among military helicopter pilots and crew.

METHODS

The study was conducted in a laboratory setting at University of Southern Denmark, Odense. A group of 11 healthy males (25,9±1,4 years) with no pain in neck or shoulders (VAS=0) were included in the project.
Electromyographic activity and fatigue development was measured in trapezius descendens, sternocleidomastoideus and the upper neck extensors. Measurements were obtained during cervical extension, cervical flexion, lateral flexion, cervical rotation, reverse flyers and shrugs. Testing was conducted over three separate days. Familiarization with the exercises and assessment of loadings corresponding to 12RM and 20RM were found on two individual days. These loadings were then used in a test session and EMG data were obtained. The order of exercises was randomized for each individual. Electromyographic amplitude during exercise testing was for each muscle offline normalized to the maximum amplitude obtained during muscle specific maximal voluntary isometric contraction (% MVE).

RESULTS

Shrugs and reverse flyers induced >60% MVE in trapezius and were significantly higher than all the remaining exercises. Trapezius activity in the remaining exercises did not exceed 20% MVE. Cervical extension produced the highest numerical activity in the upper neck extensors, but was not significantly higher than shrugs and reverse flyers during 12RM and 20RM. Cervical flexion and shrugs produced an activity >60% MVE during 12RM. Cervical flexion produced the highest activity in sternocleidomastoideus during both 12RM and 20RM but this was not significantly higher than for cervical rotation and lateral flexion. No exercises induced an activity >60% MVE in sternocleidomastoideus.
Significantly higher activity was observed during 12RM compared with 20RM during all exercises except in cervical flexion. Differences between the 12 and 20 RM were only observed for the highest activated muscles during each exercise.
A significantly higher activity was obtained during the concentric phase compared with the eccentric phase of muscle contraction.

DISCUSSION

Trapezius was specifically activated by shrugs and reverse flyers. An important observation was that reverse flyers and shrugs also induced high activity of the upper neck extensors and this was not significantly different from the specific neck exercise, cervical extension. To our knowledge this has not previously been demonstrated for the shrug exercise. This finding can have important implications as it indicate that the use of specific exercises for the cervical musculature may not be needed. Several exercises activated sternocleidomastoideus between 38%-49% MVE indicating that this muscle has multiple functions of the cervical spine. Nevertheless, no exercise induced an activity >60% MVE of sternocleidomastoideus. The concentric phase induced significantly higher activity compared with the eccentric phase of muscle contraction, indicating that the participants followed the stated repetition speed.

CONCLUSIONS

This electromyographic evaluation of neck and shoulder exercises documents that the type and intensity of exercise played a significant role for the activity in specific muscle groups. Shrugs and flyers were highly effective exercises as they induced high muscle activity of both shoulder and neck muscles. These findings have practical implications for the choice of exercises for training programs targeting muscle specific pain and disorders.

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MUSCLE COORDINATION DURING BENCH PRESS IS CHANGED AFTER 5 WEEKS OF STRENGTH TRAINING – A RANDOMISED CONTROLLED STUDY

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INTRODUCTION

It is well documented that during the initial phase of a strength training program, strength increases primarily due to neural adaptations within the nervous system. The aim of this randomised controlled study was to assess training induced adaptations in muscle synergies during bench press.

METHODS

Thirty untrained subjects were randomly allocated to either a training (TRA) (age 25.6±4.9 years, height 180.0±6.6 cm, body mass 77.2±16.2 kg) or control group (CON) (age 22.9±2.7 years, height 180.4±7.9 cm, body mass 77.2±11.1 kg). The pre- and post-test were separated by 5 weeks and consisted of performing 3 sets of 8 repetitions at 60% of 3 repetition maximum (3RM) in bench press. In between test sessions, TRA performed supervised upper body strength training 3 times a week, while CON did not train. Muscle synergies were extracted from surface electromyographic data of 13 muscles using non-negative matrix factorization [1]. A muscle synergy is defined to consist of a synergy activation coefficient and a muscle synergy vector. The synergy activation coefficient represents the relative contribution of the muscle synergy to the overall muscle activity pattern, while the muscle synergy vector represents the relative weighting of each muscle within each synergy. To evaluate changes in muscle synergy vectors and synergy activation coefficients, we performed a cross-validation analysis in agreement with [1]. In this iterative procedure, the muscle synergy vectors extracted in the pre-test were recomputed, using the fixed synergy activation coefficients from the post-test and vice versa. A similar procedure was performed for the muscle synergy vectors.

RESULTS AND DISCUSSION

Following training, TRA became significantly stronger (3RM Pre = 56.5±19.6 kg, 3RM Post = 65.7±19.7 kg, $p \leq 0.001$), while no changes had occurred in CON (3RM Pre = 55.2±12.6 kg, 3RM Post = 55.0±12.2 kg, $p = 0.546$). Two muscle synergies accounted for >90% of the overall data. The similarities between muscle synergy vector 1, 2, and synergy activation coefficient 2 for pre and post was significantly lower than the similarity between the recomputed and original synergy parameters in TRA ($p \leq 0.05$), but not in CON. The modulation of muscle synergies during bench press by a 5 week strength training program can mostly be explained by modifications in efferent neural drive.

CONCLUSIONS

The present findings indicate that strength training elicited significant changes in muscle synergies concomitant to the increase in strength.

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CLASSIFICATION OF FOOT STRIKE PATTERN WITH INCREASING RUNNING SPEED

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INTRODUCTION

Foot strike patterns in running are closely related to the running style which again influences the likelihood of injury caused by running [1]. The spontaneous foot strike pattern (SFSP) used by the runner depends also among other things on the speed of locomotion and SFSP has so far only been analyzed at a relative narrow range of discrete running speeds [2]. Therefore, to understand in detail how running velocity influences SFSP in individual runners, it is necessary to analyze changes in foot strike pattern more detailed, e.g. during continuous changes in running speed [3].

The purpose of this study was, to develop a method to determine the relationship between running speed and SFSP.

METHODS

Nine moderately trained male runners participated. Each participants completed a run, of maximal nine minutes, on a treadmill. The speed started at 4 km hr^{-1} for 1 min (walking) and was then increased with 0.5 km hr^{-1} every 15 second, for a max speed of 20 km hr^{-1}. Four foot-switches were placed in the right shoe of each runner and data were collected continuously during the test.

RESULTS AND DISCUSSION

The analysis was made by a semi-automatic step-by-step analysis of the foot switch signals. Five classes of foot strike patterns emerged from this analysis, in Table 1.

Table 1 Classification of foot strike pattern.

Heel (H)	**Running pattern where the heel is clearly activated first. More than 10 ms until the next footswitch is activated.**
HindFoot – ForeFoot (HF-FOF)	Heel and midfoot is coincide. Next, the forefoot is activated by more than 10 ms after the hindfoot.
FullFoot (FUF)	The distance between midfoot and forefoot, and between heel and midfoot, is coincide.
MidFoot-ForeFoot (MF-FOF)	The heel is not activated.
ForeFoot (FOF)	The forefoot is activated first and the heel is not activated.

Two foot-switches are categorized as coinciding, if the spacing is less than 10 ms.
The individual transitions between foot strike patterns of the participants with increasing running speed are presented in Table 2.

Table 2 The speed at which the foot strike pattern changed (km hr^{-1}). In the walk-run transition all participants perform hell strike.

	Walk-run transition	HF-FOF	FUF	MF-FOF	FOF	End time	Tend to FOF
P1	8.0	-	-	-	-	18.0	
P2	6.0	-	-	14.5	-	17.5	x
P3	7.0-7.5	-	8.0	-	11.5	20.0	
P4	7.0	13.0	19.0	-	-	20.0	
P5	8.0	8.5	12.0	19.5	-	20.0	x
P6	6.5	6.5	8.5	17.0	-	20.0	x
P7	8.0	9.0	-	17.0	-	19.0	x
P8	7.5	-	-	19.5	-	20.0	x

It is seen that the foot strike patterns change with increasing speed and that the pattern of changes are individual i.e. not systematic.

With the method used in this study, an extended classification of foot strike patterns appeared including five classes, instead of the common three (Heel, MidFoot and Toe). At the same time, there was an indication of a gradual change from foot strikes on the posterior part of the foot to foot strikes on the anterior part of the foot with increasing running speed.

CONCLUSION

The spontaneous foot strike pattern in moderately trained runners is dependent on the running speed and the speed at which the transitions between foot strike patterns occur seem to be highly individual. This indicates that more detailed analyses of intraindividual variations in foot strike patterns, in relation to variations in running speed during running training, could be helpful in the process of getting knowledge about the mechanisms behind the development of running injuries.

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DESIGN OF A SUBJECT-SPECIFIC AMERICA'S CUP GRINDING HANDLE: FROM 3D-SCANNING TO 3D-PRINTING TECHNIQUES – A TECHNICAL NOTE

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INTRODUCTION

The most physical demanding job in Americas Cup racing is grinding. Grinding is an upper body exercise similar to arm cranking. Observations have shown that grinders suffer from premature forearm fatigue. Forearm fatigue can, in some cases, lead to injury and this connection has been found in other upper body sports such as rowing [1]. Forearm fatigue and injuries also occur in grinding [2]. The handle diameter affects muscle recruitment [4], therefore custom handles might reduce forearm fatigue [3]. The aim of this study was to develop a method using 3D-scanning and 3D-printing to manufacture subject-specific grinding handles (CH), reducing forearm muscle fatigue, and test it against a standard cylidrical handle (SH).

METHODS

Three subjects participated in this randomized crossover study. The subjects performed backwards grinding at a fixed torque resistance (10 Nm) until a cadence of 120 RPM could no longer be maintained, the equivalent to 128 Watt. This experiment consisted of two trials, one with SH and one with CH, performed on separate days. Subjects' maximal grip force (MVC) was measured with a hand-dynamometer before (baseline) and after (post) grinding. The median power frequency was calculated for the first and last 20 s using a Short Time Fourier Transform (window length of 10 s) in MATLAB® (Mathworks, Massachusetts, USA). The sample rate was set to 2000 Hz. The measured muscles were: the extensor carpi radialis longus (ECR), the flexor carpi ulnaris (FCU) and the flexor digitorum superficialis (FDS). A custom-made cylinder with a varying diameter along its axis (handle core) was first created, as shown in Figures 1-A and 2-A. When comparing a standard cylindrical handle (Figure 1-B) with the handle core (Figure 1-C), it is possible to note that the index, ring and little fingers showed better alignment with the middle finger in the latter case. Theoretically, this corresponds to a higher absolute grip strength since each finger works at a proper length [5].

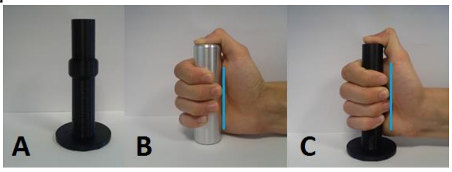

Fig. 1 A) Handle core; B) Griping around a 32mm cylindrical handle; C) Griping around the handle core.

Later, in order to create the custom-fitted handle, a thin layer of plasticine (play-doh) was wrapped around the handle core (Figure 2-A). The subjects were asked to squeeze the play-doh around the handle until they could feel the cylinder core (Figure 2-B). Once the handle core was released, a final handprint was engraved in the play-doh. The engraved handprint was later 3D-scanned using a MakerBot Digitizer turntable (MakerBot Industries, New York City, USA). The mesh was then imported to Sculptris® software v.Alpha-6 (Pixologic Inc., California, USA) where the raw surface was smoothed. The surface mesh was, hereafter, imported into SolidWorks® CAD software (Dassault Systèmes SolidWorks Corp., Massachusetts, USA) and the surface mesh was converted into a volume mesh and, consequently, to a solid geometry. Finally the respective custom-fitted handle was 3D-printed.

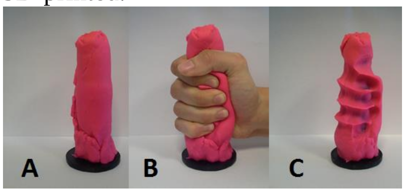

Fig. 2 Three steps illustrating the development of the customized handles: A) Plasticine wrapping; B) Subject gripping around the play-doh; C) handprint.

RESULTS AND DISCUSSION

Descriptive statistics (median and interquartile ranges) were chosen as results due to small sample size. The time to fatigue was slightly longer for CH (173 s, 148 to 190) than SH (170 s, 144 to 188). The reduction in MVC was also lower for CH (12.7%, 11.4 to 14.6) than SH (14.6%, 11.8 to 15.8). A drop in the median frequency occurred for both handle types as it can be confirmed in Table 1.

A comparison test between CH and SH did not show a large difference between handle types in both time to fatigue and MVC, but the CH seems to perform slightly better. However, a larger sample size is required to confirm this trend. The results indicate that the drop in median frequency over time was lower with CH, but once again, a larger sample is required to assess whether statistical significance exists.

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Table 1: Median power frequency for ECR, FCU and FDS

Muscle	Handle type	Median (%)	Q1-Q2 range
ECR	SH	6.54	6.48 to 6.68
	CH	5.19	3.09 to 8.17
FCU	SH	7.51	3.51 to 8.02
	CH	0.75	0.30 to 3.30
FDS	SH	8.19	2.07 to 15.93
	CH	4.62	0.59 to 5.53

Prediction of ground reaction forces and moments during running

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INTRODUCTION

In inverse dynamic analysis (IDA) of musculoskeletal models, predicting ground reaction forces and moments (GRF&Ms) can reduce dynamic inconsistency and obviate the need for force plate (FP) measurements. The latter would be particularly advantageous for sports science research, because it can be difficult to ensure FP impact during movements that are highly dynamic and require a large space. Therefore, this study aimed to validate the method of Fluit et al. [1] to predict GRF&Ms during running by comparing the predicted variables and associated joint kinetics to the corresponding variables from a model, in which the GRF&Ms were measured using FPs.

METHODS

Ten healthy subjects were instructed to run at a comfortable self-selected pace and impact the FP with their right foot. 35 reflective markers were placed on the subjects (29 markers on the skin surface and three markers on each running shoe) and their trajectories tracked using eight infrared high-speed cameras (Oqus 300 series), sampling at 250 Hz, combined with Qualisys Track Manager v. 2.9 (Qualisys, Gothenburg, Sweden). GRF&Ms were obtained at 2000 Hz using two FPs (AMTI, Inc., Watertown, MA, US), which were embedded in the laboratory floor. Three out of five successful trials for each subject were included for further analysis.

Musculoskeletal models were developed in the AnyBody Modeling System v. 6.0.4 (AnyBody Technology A/S, Aalborg, Denmark) based on the *GaitFullBody* template from the AnyBody Managed Model Repository v. 1.6.3. Model scaling and kinematic analysis were performed applying the methods of Andersen et al. [2,3]. Constant strength muscles were added to the lower extremities and the muscle recruitment problem was solved by minimizing the sum of the squared muscle activities. The GRF&Ms were predicted by creating five artificial muscle-like actuators at 18 contact points under each foot of the musculoskeletal model. One actuator was aligned with the vertical axis of the FPs and generated a normal force, while the other actuators were defined in two pairs aligned with the medio-lateral and antero-posterior axis of the FPs, generating positive and negative static friction forces. A dynamic contact model was implemented to determine foot-ground contact, ensuring that each actuator would only be activated if their associated contact point was sufficiently close to the floor and almost without motion. The magnitudes of the predicted GRF&Ms, i.e. the activation level of each actuator, were solved as part of the muscle recruitment algorithm, hereby, providing a solution to the problem of under-determinacy during the double support phases.

Data were analyzed from the first foot-FP contact instant to the last frame of contact. Vertical GRF, sagittal GRM, ankle, knee and hip resultant joint reaction force (JRF) were compared using Pearson's correlation coefficient and root-mean-square deviation.

RESULTS AND DISCUSSION

For all selected variables, comparable results were obtained between the two datasets (Table 1), highlighted by the excellent correlation for the vertical GRF (Figure 1).

This study validated a method to predict GRF&Ms from full-body motion only during running, which provided comparable results to traditional IDA for all analyzed variables. Based on these results, this method could be used instead of FP measurements, hereby, facilitating analysis of sports-related movements and enabling complete IDA using systems that do not provide an interface between kinematic and FP data.

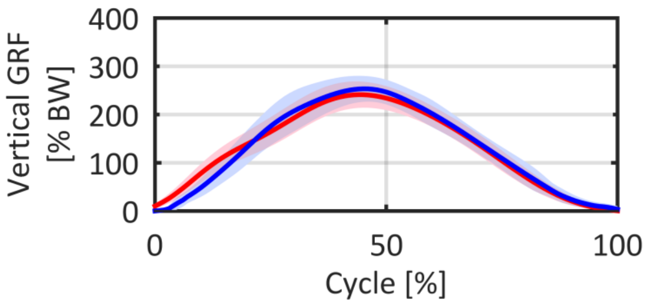

Fig. 1 Vertical GRF during running. The predicted (blue) and measured (red) variables presented as the mean ± 1 SD (shaded area) over all trials.

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Table 1 Pearson's correlation coefficient (*r*) and root-mean-square deviation (RMSD) for selected variables during running.

	Vertical GRF	Sagittal GRM	Ankle resultant JRF	Knee resultant JRF	Hip resultant JRF
r	0.99 ± 0.00	0.87 ± 0.09	0.93 ± 0.04	0.98 ± 0.01	0.94 ± 0.05
RMSD	15.09 ± 3.45	3.59 ± 1.50	177.49 ± 63.00	74.92 ± 22.47	100.31 ± 23.37

Kinematic And Kinetic Effects Of Variable Resistance On The Conventional Deadlift

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INTRODUCTION

The use of chains in conjunction with submaximal loads in the deadlift is a common practice among powerlifters and strongmen to develop strength and power[1]. Chains increase the resistance in a linear fashion as chains unfold from the floor. Adding chains to the barbell is thought to suit the ascending strenghcurve of the deadlift.

Anecdotal evidence testifies to the effect of chains, Swinton et al.[3] has though performed the only analysis of the conventional deadlift with the inclusion of chains. In this study the subjects where permitted to elevate their heels, which is not allowed in powerlifting competitions [2]. Thus, the aim of the present study was to investigate the kinematic and kinetic effects of different barbell- and chain loads on the conventional deadlift performed arcording to competition rules.

METHODS

Ten experienced resistance trained men volunteered to participate in this study (deadlift 1RM:227±25 kg; resistance training experience: 7.6±3.97 years).

Data were collected for each subject over two sessions separated by one week to ensure sufficient recovery. Session 1 involved 1RM testing in the conventional deadlift and session 2 involved a total of 16 submaximal trials at 40, 50, 60 and 70% of 1RM across four chain conditions (No chain, 15%, 25% and 35% of 1RM). Chains were attached so that the average load lifted in the chain and no chain conditions were equated by subtracting half the mass of the chains from the initial barbell load. For example at the 25% chain condition, the barbell load was 12,5% less at the bottom, equal at half and 12,5% greater at the top then the no chain condition.

All trials were performed on a force plate and a stringpot potentiometer was attached to the barbell to measure vertical displacement. Barbell displacement, vertical ground reaction force and duration of the concentric action were measured. Velocity, power and relative duration of the acceleration phase were calculated.

RESULTS AND DISCUSSION

A MANOVA test showed significant effect of barbell and chain load on all variables displayed in *Table 1*. The relative effect of chain conditions became less pronounces as barbell load increased and more pronounce as the magnitude of chain resistance increased. Utilizing chains decreased all the dependent variables shown in table 1 but enabled greater force to be maintained throughout the lift (*Fig. 1*).

The results are generally in accordance with those of Swinton et al. [3].However, despite the use of an identical loading protocol Swinton et al.[3] reported increased F_{peak} with chain resistance. This discrepancy might be related to the ankle joint extension permitted in Swinton's trials.

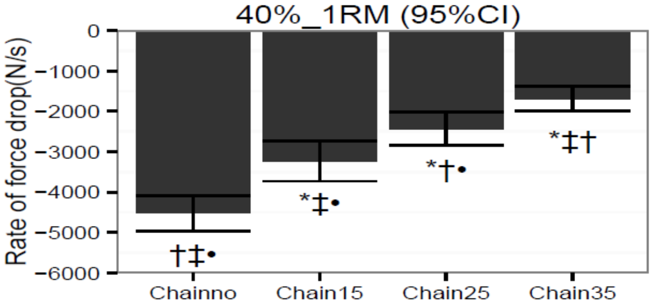

Fig. 1 Rate of force drop for intensity 40%, see table 1 for symbol description.

CONCLUSIONS

The results of the current study suggest that chain resistance may be advantageous tool for coaches and athletes looking to reduce the rate of force drop in submaximal deadlifts.

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	F_{peak}(N)	$V_{average}$($m\times s^{-1}$)	V_{peak}($m\times s^{-1}$)	$P_{average}$(W)	P_{peak}(W)	a_{phase}(%duration)
40% $_{Chainno}$	2618 ± 263‡•	1.01 ± 0.07†‡•	1.56 ± 0.11†‡•	1716 ± 251‡•	3204 ± 401‡•	55 ± 6‡•
40% $_{Chain15}$	2558 ± 258•	0.94 ± 0.07*•	1.43 ± 0.11*•	1683 ± 231•	3005 ± 373•	47 ± 4•
40% $_{Chain25}$	2499 ± 261*•	0.88 ± 0.05*	1.37 ± 0.09*•	1641 ± 223*	2893 ± 350*•	43 ± 5*•
40% $_{Chain35}$	2448 ± 255*†‡	0.83 ± 0.07*†	1.29 ± 0.11*†‡	1610 ± 238*†	2840 ± 434*†‡	40 ± 7*†‡

Table 1 In the table we see the values for 40% of 1rm at different chain loadings.

* Significant difference between value and **40%** $_{Chainno}$ ($p<0.05$) † Significant difference between value and **40%** $_{Chain15}$ ($p<0.05$)

‡ Significant difference between value and **40%** $_{Chain25}$ ($p<0.05$) • Significant difference between value and **40%** $_{Chain35}$ ($p<0.05$)

DEVELOPMENT OF A WORKFLOW FOR WEAR AND FUNCTIONAL SIMULATION OF TOTAL KNEE ARTHROPLASTY

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INTRODUCTION

Total knee arthroplasty (TKA) is a successful treatment for endstage osteoarthritis. Although several advancements in implant design and materal have emerged over the years, bearing wear still remains a limiting factor for the longevity of the prostesis, especially for young and active patients [1].

In the past years, several computer models, simulating experimental knee wear have been developed to replace costly wear experiments. However, these are controlled by standardised loads in only a few degrees-of-freedom (DOF) and thus do not represent the in-vivo knee joint load accurately. The aim of this project was therefore to develop a workflow for simulating the patient-specific wear. This was accomplished using musculoskeletal simulation in the AnyBody Modeling System (AMS) (AnyBody Technology A/S, Denmark) to estimate the body kinematics, joint and muscles forces that were subsequently applied to a finite element (FE) model in FEBio (University of Utah & Columbia's Musculoskeletal Biomechanics Laboratory). The FE model contained the tibial and femoral implants, and the major surrounding ligaments, and was solved in all six DOFs. Based on the FE results, the wear was estimated for the tibial implant using Archard's linear wear model.

To investigate the effect on estimated wear, two different knee models were used in the AMS model 1) a hinge model assuming only flexion-extention movement of the knee and 2) a contact-based model, using force-dependent kinematics (FDK), which is computationally slower than the hinge model but allows a more accurate joint model.

METHODS

The AMS model was based on the *GaitLowerExtremity* model from the AnyBody Managed Model Repository (AMMR) 1.3. The two implants, femoral component made of titanium and tibial component of polyethylene, were positioned based on a CT scan from the first Grand Challenge Competition in 2010 [2]. Those data also included motion capture for a gait cycle from which the kinematics were estimated in AMS.

Loads and kinematics were exported and applied to the FE model in which the polyethylene was modeled as a hyperelastic material with a volumetric 4-noded tetrahedral mesh and the femoral component was rigid with 3-noded triangular shell elements. The flexion-extension angle was displacement-controlled while the other five DOFs were force-controlled. The included ligaments were modelled as nonlinear springs with properties based on [3].

The frictionless contact problem was solved using a penalty method with a penalty factor of 20. The rigid object (femoral component) was chosen to be the master.

The computed kinematics and contact pressure between the two implants, were applied to the wear model with a wear coefficient defined as 2.64 10^{-10} $mm^3N^{-1}mm^{-1}$ [4].

RESULTS AND DISCUSSION

The obtained wear contour after one gait cycle for the two AMS models is shown in Figure 1.

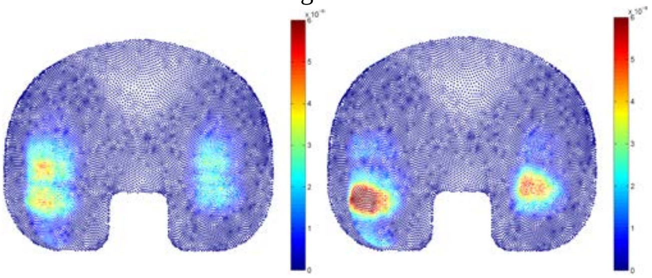

Fig. 1 Wear contour of the tibial implant (medial side to the left) after one gait cycle in 10^{-6} mm of the hinge joint model (left picture) and the FDK model (right picture).

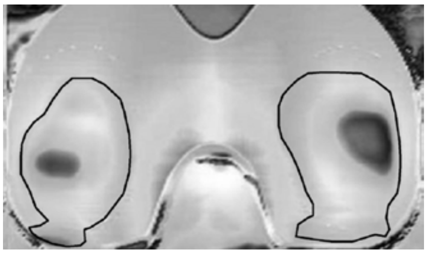

Fig. 2 Experimental surface profilometry for the tibial implant (medial side to the right) after 5 million cycles [4].

Both vary from the typical results from wear simulators (example shown in Figure 2), which can indicate the lack of in vivo representation for the latter method. However, this is still to be further investigated with in vivo validations.

CONCLUSIONS

The developed workflow of going from patient-specific movement and geometric data to wear estimations of knee implants, is considered successful even though the FE model was only solved for 65% due to an unknown error.

The obtained wear results require validation against in vivo measurements from TKA patients instead of wear simulator results.

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BORN TO JUMP – MOTION PREDICTION USING FORWARD DYNAMICS

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INTRODUCTION

In several cases of human movement, different people appear to display nearly identical kinematics. During maximum height squat jumping, different subjects prefer different body configurations at the start of the jump. During the jump, however, the joint angle histories of the subjects converge to a common, stereotyped pattern [1]. So why do people jump the way they do? But maybe even more interestingly - do people jump optimally? And might some people have an inherent talent for jumping?

Specific and focused training may naturally increase the resulting joint torque capabilities around the ankle, knee, and hip, resulting in an increased jumping height when properly coordinated. But what about the significance of aspects that are not possible to adapt, such as the inherent anthropometry of the lower extremities?

This preliminary work aims at revealing the optimal lower extremity composition for maximum height squat jumping for otherwise identical people with respect to total mass, total height, and joint torque strengths. Results rely on optimal motion prediction of a mathematical model of a skeletal system with joint torque actuators. The joint torque stimulation of a planar forward dynamics multibody model is optimized to maximize jump height.

The non-linear nature of the multibody dynamics (MBD) system introduces several local minima to the non-convex design space. For this reason, global zeroth order search algorithms must be applied. The optimization algorithm Simulated Annealing has been found to work well for a small to moderate number of design variables.

METHODS

The skeletal system with joint actuators considered in this work is modeled as four rigid segments, interconnected by three revolute joints, resulting in six degrees of freedom. Inputs for the forward dynamics model are the time histories of the joint torque stimulations, and the resulting motion of the skeletal system is simulated by numerical integration.

For now, joint torque strengths are implemented as Hill-type inspired functions based entirely on the isometric angular position dependencies extracted from experimental data collections of a single sprinter. The isokinetic strength dependency is work in progress.

The joint torque stimulations for each of the ankle, knee, and hip joints are allowed to vary linearly in between five control points equally spaced in time, resulting in a total of 15 design variables to be optimized. The objective function to be maximized is out of stability considerations defined as the maximum vertical coordinate position of the head center throughout the simulation time, set to 1.50 seconds.

The initial configuration (t = 0.00 s) of the planar MBD model appears in Fig. 1 (left).

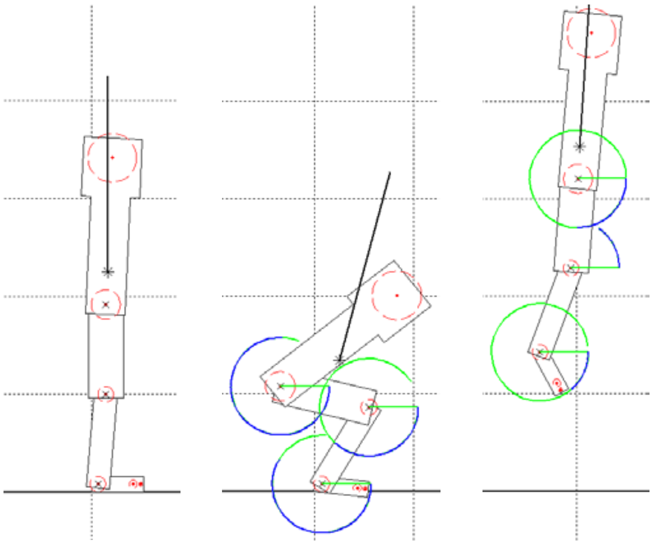

Fig. 1 Initial (left), intermediate (mid), and maximum height (right) simulation instances of the MBD model. Dotted red circles are contact spheres for ground (horizontal black line) contact modeling. The black star and the attached line are the center of mass and angular orientation of the system, respectively. Circular arcs in green and blue represent joint torque stimulation and actual strength fractions, respectively.

The total mass and height of the considered skeletal model are fixed at 87 kg and 1.85 m, respectively. The sizes of the lower extremity body parts are, however, varied within an assumed natural variation of as much as +/- 10% compared to standardized segment proportions [2]. Joint torque strength profiles are assumed identical for all of the variations.

RESULTS AND DISCUSSION

An intermediate (t = 0.56 s) and the maximal height instance (t = 1.19 s) of an optimization procedure appears in Fig. 1 (mid) and (right), respectively, with an objective function value of 2.35 m, corresponding to a vertical coordinate position increase of 65 cm compared to the initial configuration.

As joint torque strengths, segment masses, and inertia properties are not yet fully implemented, reliable conclusions for this promising albeit preliminary work remain to be done.

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DOES KINESIOTAPE FACILITATE OR INHIBIT THE ACTIVATION OF THE TRAPEZIUS MUSCLE?

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INTRODUCTION

Shoulder injuries are common in sports and occupations requiring repetitive arm movement and movement above the shoulder level. Kinesiotape has been suggested for the prevention or treatment of musculoskeletal injuries [1]. However, the effects of scapular taping on the surface electromyographic (SEMG) activity of the shoulder muscle are still subject to debate [1,2].

The purpose of this study was to examine the effect of kinesiotape on SEMG activity of the upper (UT) and lower (LT) trapezius during a static and dynamic exercise.

METHODS

Eight healthy badminton players participated in the study (age 21.1±2.2 years, height 1.8±0.1 m, body mass 75.3±8.7 kg, body-mass index 22.3±1.7 kg/m^2). The static exercise consisted of an abduction of 30° anterior in the frontal plane, while the dynamic exercise was abductions 30° anterior in the frontal plane at a rhythm of 20 beats per minute (Figure 1). All subjects performed both contractions with and without kinesiotape (randomized balanced order). SEMGs were recorded from the UT and LT muscle. Root mean square (RMS) and normalized mutual information (NMI) values were computed to estimate the level of muscle activity (normalized with respect to a reference contraction) and the level of functional connectivity among the UT-LT muscle pair, respectively [3]. A Wilcoxon signed rank test was used. Median [25-75 %] quartiles are reported. $P < 0.05$ was considered significant.

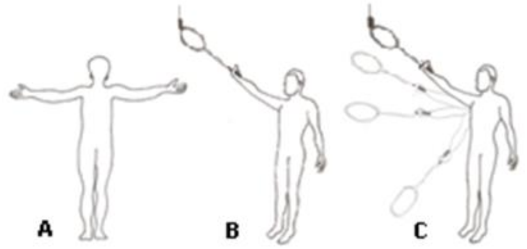

Figure 1: Reference contraction (A), static contraction (B) and dynamics contraction (C)

RESULTS AND DISCUSSION

There were no significant differences in normalized RMS values of the UT and LT trapezius during the static exercise. On the contrary, the normalized RMS values of the UT trapezius were significantly lower with kinesiotape compared with without kinesiotape during the dynamic exercise, i.e., 250.3 [153.3-331.3] vs. 317.1 [165.6-291.0] ($P < 0.05$). Further, the NMI values (Figure 2) of the UT/LT trapezius were significantly lower with kinesiotape compared with without kinesiotape during the static exercise, i.e., 0.022 [0.017-0.023] vs. 0.023 [0.019-0.024] ($P < 0.05$). There were no significant differences in NMI values during the dynamic exercise.

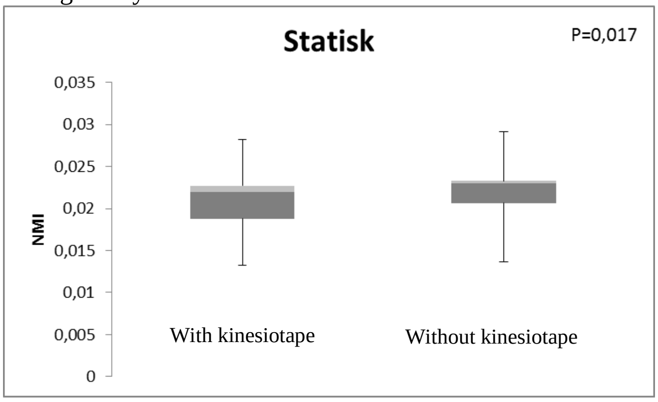

Figure 2: Median [25-75 %] quartiles normalized mutual information (NMI) values of the upper and lower trapezius muscle pair during static contraction with and without kinesiotape.

CONCLUSIONS

The present results indicate that kinesiotape applied across the muscle fibers of upper trapezius can reduce the muscle activity during dynamic exercise and the functional connectivity of the UT/LT during static contraction. Further studies are needed to confirm these findings.

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VARIABILITY PATTERN OF FORWARD BENDING OF THE TRUNK AMONG BLUE-COLLAR WORKERS

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INTRODUCTION

The quantification of variation in relation to exposure is important in the analysis of occupational biomechanics [1] and should be considered for both work and leisure [2]. For this purpose Exposure Variation Analysis (EVA) has been suggested as a computational framework to quantify variation at work [1]. The principal function of EVA is to simplify the extent of intricate time-line of exposure on both pre-defined exposure category levels and duration of uninterrupted sequence pre-defined categories, and combine these into a 2-dimensional matrix. The purpose of this study was to develop and apply EVA on diurnal recordings of physical activity focusing on forward bending during both work and leisure.

METHODS

This studied population is based on a subsample of male manufacturing workers (n=32 males, mean age±SD 26.1±4.0 years) with similar duration of working (7.9±0.9 h) and leisure hours (6.1±0.8 h) during a working day from the field study called 'Danish Physical ACTivity cohort with Objective measurements (Dphacto)' [3]. Two triaxial accelerometers (ActiGraph GT3X+, ActiGraph LLC, Pensacola, FL, USA) were used to detect trunk inclination and body posture (Figure 1).

In this study all samples of the selected periods of trunk inclination, with respect to upright standing posture as the reference position, were categorized in intervals of [<5, 5-10, 10-20, 20-40, 40-80 and >80 deg.] and time categories of [0-3, 3-7, 7-15, 15-31 and ≥31 s]. This resulted in matrixes in which each element represents an accumulated elapsed time that the trunk inclination stays uninterruptedly in an interval for the determined time categories. Thus, 5 levels along the time axis and six levels along the amplitude axis were generated. The centroid of the map 5x6 plane was extracted, providing information about the general tendency towards a displacement of exposure into the determined intervals and time categories [4]. Additionally, the standard deviation of the elements of EVA matrix was calculated.

A repeated measured ANOVA was used to reveal the difference of exposure profiles in work and leisure time activity of the manufacturing workers in terms of i) the standard deviation of the elements of EVA matrix, ii) the centroid of the EVA layout along the time and the amplitude.

RESULTS AND DISCUSSION

The standard deviation of the elements of EVA matrix during work was significantly larger than during leisure, 510.4±201.4 s vs. 158.7±79.8 s, respectively (F=78.966, p<0.001). This indicates that the manufacturing workers had a more variable trunk inclination pattern during leisure compared to work. Similarly, the centroid of the EVA profile was significantly shifted towards smaller uninterrupted time categories during leisure compared to work, 2.2±0.3 vs. 2.4±0.4, respectively (F=13.023, p=0.001), indicating that the workers stayed with the trunk forward bended for shorter duration of time during leisure.

Having the contrasting health outcomes of physical activity (i.e. forward bending) at work and leisure in mind, these findings supports previous suggestions that a more variable pattern of exposure may be beneficial when addressing forward bending in relation to the intensity of low back pain [2]. This implies that studies regarding forward bending should not only focus on the duration of forward bending, but also the variability pattern of exposure in relation to the intensity of low back pain [2].

Fig. 1 Accelerometer setup for detecting forward bending of the trunk (accelerometer on T1-T2) in standing posture (accelerometer on thigh).

CONCLUSIONS

This study revealed for the first time that the EVA enables detection of differences in physical exposure during work and leisure. That is, the physical exposure pattern during leisure time is more variable than the exposure pattern at work.

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NEURO-MECHANICS OF STRETCH-SHORTENING MUSCLE ACTIONS IN LOCOMOTION MOVEMENTS

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Through evolution two-legged walking, running and locomotor-like movements (hopping and skipping) have emerged, as movement patterns that are executed spontaneously, when humans move. A common characteristic of these movement patterns is the stretch-shortening muscle action, that the larger extensor muscles of the lower extremities, undergoes, in the movement phase where the foot or feet are in contact with the ground.

Stretch-shortening cycles (SCC's) of muscle actions are highly energy efficient, when they are coordinated optimally. The economy of movements dominated by SSC's, and the mechanically efficiency often exceeds the theoretical muscle mechanical efficiency of about 25%. This may seem like a paradox, but it is possible, because muscles are combinations of muscle fibers in series with tendinous structures, and the active muscle-tendon unit has elastic properties dominated by the elastic properties of the tendinous structures. During cyclic SSC's kinetic energy is temporally stored the tendons as elastic energy during the stretch and this energy is released again during the shortening, which reduces the metabolic cost the muscle fiber actions.

Optimal coordination of SCC's implicates that the burst of muscle activity is timed in such a way that the muscle fiber yielding (eccentric muscle action) is minimized, and the time between the end of the stretch and the start of the shortening is as short as possible. The activity patterns in the motoneurone pools of the muscles controlling the SSC's during locomotion and locomotor-like movements is generated by a combination of neural inputs from supra-spinal centers in the brain and the brain stem, networks of interneurons in the spinal cord and afferent input. The relative contributions of the inputs to the motoneurone pools during locomotor movements, the cycle-to-cycle modulation of these inputs and the effect of acute and prolonged exercise and training are still not fully clarified.

A thorough understanding of the neuro-mechanical organization of stretch-shortening muscle actions during human physical activities and the influence of e.g. exercise, training, immobilization and genetic predisposition is important for improvement of performance in rehabilitation and sports.

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