Less hip joint loading only during running rather than walking in elderly compared to young adults

Georgios Giarmatzis¹, Ilse Jonkers², Remco Baggen³, Sabine Verschueren^{1,*}

¹ Research Group for Musculoskeletal Rehabilitation, Department of Rehabilitation Sciences, Katholieke Universiteit Leuven, Tervuursevest 101, B-3001 Heverlee, Belgium

² Human Movement Biomechanics Research Group, Department of Biomedical Kinesiology, Katholieke Universiteit Leuven, Tervuursevest 101, B-3001 Heverlee, Belgium

³ Physical Activity, Sports & Health Research Group, Department of Rehabilitation Sciences, Katholieke Universiteit Leuven, Tervuursevest 101, B-3001 Heverlee, Belgium

* Corresponding author: Georgios Giarmatzis, Research Group for Musculoskeletal Rehabilitation, Department of Rehabilitation Sciences, Katholieke Universiteit Leuven, Tervuursevest 101, B-3001 Heverlee, Belgium. giorgos.giarmatzis@faber.kuleuven.be

Acknowledgements

The authors would like to thank all the participants in this study for their outstanding cooperation.

This study was supported by grant G0526512 from the Fund for Scientific Research (FWO-Vlaanderen) to S. Verschueren.

22

23 Abstract:

24 Walking and running have been found to increase hip bone mass in postmenopausal women.
25 However, the optimal speed to trigger osteogenesis is still under debate because the exact
26 loading during different speeds is poorly characterized. Moreover, age related differences in
27 gait kinematics/kinetics can potentially result in differences in peak hip loading, making
28 extrapolation of results based on young populations to the elderly misleading. Using
29 integrated 3D motion capture and musculoskeletal modeling, peak hip contact forces (HCFs)
30 were calculated during walking and running from 3 to 9 km/h in 14 female young (21.4 ± 1.6
31 years old) and elderly (69.8 ± 3.4 years old) participants. Peak HCFs were similar during
32 walking in both groups, whereas elderly loaded their hip less than young during running,
33 through reducing their stride length and hip adduction angle at peak loading. Moreover, hip
34 adduction moment was found to best predict peak HCF during impact in walking and running
35 whereas hip extension and external rotation moment can predict the second peak HCF during
36 walking in the elderly and young group respectively. Comparison between same speeds in
37 walking and running revealed that in contrast to young no additional hip loading is imposed
38 during running in elderly. The present study offers an insight into the differences in hip
39 loading profile in postmenopausal women during walking and running at different speeds.
40 Such information is crucial to medical experts that target site-specific bone loading through
41 exercise in elderly populations in order to prevent hip bone loss.

42 **Keywords**

43 Osteoporosis, biomechanics, exercise, hip contact forces, musculoskeletal modeling, elderly

44

1. Introduction

Deterioration of bone microstructure and consequent bone fragility caused by osteoporosis constitute an important public health issue [1], urging for more efficient strategies to address it. Mechanical loading imposed by exercise promotes bone formation [2], however not all exercise regimes favorably affect targeted bone areas [3,4]. Fast walking (5-6 km/h) and running were found to promote femoral neck bone mineral density (BMD) [5,6]. However, what constitutes an optimal exercise program with regards to bone formation in the elderly is unknown [7,8] due to insufficient knowledge on the exercise-specific effect on bone loading.

Forces acting on the hip joint during exercise are responsible for local deformations and the resulting osteogenic responses inside the bone [9,10]. In-vivo joint forces are not available in healthy populations. Instead, ground reaction forces (GRFs) [11,12] have been used to estimate them, although they are considered insufficient [13]. Hence, musculoskeletal modeling offers a unique non-invasive way to calculate joint contact forces during human motion.

The purpose of this study was to explore the differences during walking and running between young and elderly in terms of peak hip loading and the related movement strategy. We hypothesized that elderly will show decreased hip loading, since they showed reduced joint kinetics [14,15] during walking and running compared to young.

2. Methods

2.1. Participants

14 healthy young (mean age: 21.4 years, mean BMI: 22.6) and 14 elderly (mean age: 69.6 years, mean BMI: 24.4) female participants were recruited for this study. Gender selection was justified as post-menopausal women are the main target group for most training programs that address osteoporosis. The study was approved by the local Ethics Committee and informed consent was obtained from all subjects. Exclusion criteria were prior conditions that might affect their performance, such as pain, lower limb fracture and surgery.

2.2. Gait analysis

An instrumented split-belt treadmill (Forcelink, Culemborg, The Netherlands) was used in this study to impose the same speeds of walking and running in all subjects, who were instructed to walk and run at speeds ranging between 3 km/h up to the highest comfortable speed, with speed increments of 1 km/h. 9 km/h was the highest attained speed for both age groups. Standardized sport footwear was provided. A body harness attached to the ceiling was used, in case of treadmill failure or fall.

All subjects were instructed to independently choose their transition speed from walking to running, look forward and move their hands freely. The number of subjects in each speed is shown at Table 1. Since most of the elderly participants had little or no experience with the treadmill, enough time was given in order to acclimate themselves with every speed. However, the acclimation period did not last more than ~1 minute, since the goal was to limit activity time and fatigue. The recording time was set for each speed to 10 seconds.

50 retro-reflective markers [13] were placed on the participants' bodies and their motion was captured using a ten-camera VICON system (10-15 MX camera system, VICON, Oxford Metrics, Oxford, UK) sampling at 100 Hz. Clustered markers were placed on the thigh and tibia to improve 3D kinematics calculations of these body segments. Medial markers at the knee and ankle joint were removed during dynamic trials. The three components of the

GRFs were measured by two force plates integrated in the treadmill sampling at 1000 Hz and filtered at 6 Hz [13].

2.3. Musculoskeletal modeling

Opensim (Stanford University, CA, USA) was used to perform musculoskeletal modeling. After scaling a generic musculoskeletal model with 23 degrees of freedom (DOF) and 92 Hill-type muscle-tendon actuators [16] to each subject's anthropometric dimensions and weight, joint angles were calculated using the Kalman smoothing algorithm [17], serving as inputs in combination with GRFs to inverse dynamics, that calculated joint kinetics. The hip joint was modelled as a 3 DOF joint, whereas the knee and ankle as 1 DOF joint. Muscle forces were calculated using the static optimization algorithm included in Opensim [18], that calculates the individual muscle forces that equilibrate the intersegmental joint moments. The cost function was set to minimize the squared sum of all muscle activations. During walking trials, muscle activation levels were restrained up to 0.8, in order to produce more physiological activation values [19]. Reserve actuators were added to the DOFs of the model, to provide the additional torques if needed during static optimization. In order to account for the reserve actuators torques in the frontal plane at high speeds (8 and 9 km/h), the hip abductor actuator moment was distributed over the different bundles of the gluteus medius force [13]. Finally, joint reaction analysis was used to calculate the three components of HCF in the femur coordinate system of the right leg. Resultant HCFs and vertical GRFs were normalized to subjects' body weight (BW) and external joint moments to subjects' body mass (Nm/kg) [13].

2.4. Statistics

Individual gait cycles were identified and all data were time normalized to 100 points by spline interpolation (Matlab 2014a, Natick, Massachusetts, United States). A first

averaging process resulted in an ensemble average curve for each subject at every speed. Then, ensemble curves of GRFs and HCFs and hip moments were calculated for all subjects along the three planes at the different speeds. A similar process was followed for the peak GRFs and HCFs. First, peak values were identified in every gait cycle for individual subjects and then first averaged to mean peak values for every subject. A second averaging process between subjects resulted in the mean peak values for every speed.

Mean right hip adduction angles at the time of the first peak during walking and running (impact peaks) were calculated, as indicator of the position of the foot and the center of pressure (COP - GRF point of application) relative to the hip joint centre.

Step length was calculated as the time between two consecutive heel strikes multiplied with the speed. It was normalized to leg length, defined as the distance between the markers placed at spina iliaca and medial malleolus in a standing position.

Linear mixed modeling was used to identify significant effects of speed and age on the measured outcomes. Speed and the interaction term of speed and age (speed x age) was inserted as fixed effect in the model in order to identify if there are any differences between the two age groups in overall and each speed during walking and running. Separate models were introduced for each group, with only speed as a fixed effect, to test the speed effect on each measured variable for each group. Subject-specific intercept was added as a random effect in all models to account for subject variability, since a separate intercept for each subject is included in the model. Significance level was set to 0.05. Stepwise regression analysis was used to identify the best predictors of peak HCFs, considering the coinciding hip moments. Regression at 7 km/h walking was not carried out due to the limited sample size.

3. Results

Peak HCFs and GRFs increased with increased speed in both groups during walking and running (Figure 1 and 2) except for HCF's in elderly during running. More specifically, first and second peak of HCFs during walking increased with increasing speed ($p < .001$) in elderly and in young ($p < .001$) (Figure 3). During running, peak HCFs did not increase with increasing speed in elderly ($p = 0.422$) but they did in young ($p = 0.002$). Moreover, first peak of GRFs increased with increasing speed ($p < .001$) in elderly and both peaks in young ($p < .001$) (Figure 3). However, second peak of GRF in elderly did not follow the speed increase ($p = 0.565$), remaining relatively constant. During running, peak GRFs increased with increasing speed in elderly ($p = 0.001$) and in young ($p < .001$). Finally, step length during walking and running also increased with increasing speed in elderly and in young ($p < .001$).

The effect of speed x age on the variables showed that differences between the two groups in terms of peak hip loading were significant only during running. Both peak HCFs were not different between age groups during walking ($p = 0.11$ and $p = 0.2$ for first and second respectively), but they were during running ($p < .001$) with lower peak HCFs in elderly in all speeds. Also, first peak GRFs were not different between age groups during walking ($p = 0.061$) except at 4 km/h, but the second peak GRF during walking ($p < .001$) and the peak GRF during running ($p < .017$) were significantly lower in elderly at all speeds and only at 8-9 km/h respectively (Figure 3). Stride length was significantly lower in elderly both during walking ($p = 0.001$) and running ($p = 0.013$). Elderly adducted their leg less than the young during the impact peaks of HCFs in all speeds of walking ($p < .001$) and running ($p < .001$).

The results of the regression analysis show that during walking, hip adduction moments is the best predictor for the first peak of HCFs in both groups at all speeds (Table 2). Hip extension moment best predicted the second peak of HCFs in elderly at all speeds except at 5 km/h. Hip external rotation moment best predicted this peak in the young group. During running, hip adduction moment was the best single predictor of the peak HCF in both groups.

No combination of moments was found to offer better prediction of peak HCFs during walking and running over all speeds.

4. Discussion

The current study calculated the forces acting at the hip of healthy elderly and young adults during walking and running. The findings suggest that elderly loaded their hip less than the young during running and the mechanisms behind this finding were explored.

The magnitude of hip joint loading calculated in the elderly subjects in the current study is consistent with in vivo data reported for THA patients during walking [20] and running [21]. Our elderly data were comparable for both walking and running to in vivo measurements with controlled speeds [21] (~ 3.5 BW at 3 km/h - ~ 4 BW at 5 km/h walking, ~ 5 BW at 7 km/h running compared to our results: 3.3 BW, 3.7 BW and 6.15 BW respectively).

We have previously established [13] that vertical peak GRFs do not reliably predict peak HCFs during walking and running. Instead, certain hip moments were found to best predict peak HCFs as they better reflect the effect of the movement pattern adopted than GRFs. The current study further strengthens this finding. The second peak of HCF increases although GRF remains relatively constant during walking in elderly. The opposite occurs during running. The observed discrepancies between peak HCFs and GRFs suggest that muscle action is an important determinant of peak loading, given the considerable magnitudes of muscle forces required to balance the external hip moments. The results of the regression analysis show that hip moments can significantly predict peak HCFs at almost every speed. Hip adduction moment was found to best predict the impact peaks during walking and running in both groups. Also, hip extension and external rotation moment best predicted the second peak during walking in elderly and young respectively.

The age related differences in peak hip loading can be related to adaptations in the movement strategies and the resulting changes in joint moments. Stride length and hip adduction angle at peak loading were reduced in elderly during walking, however the effect seems minimal, as no statistically significant differences between the peak HCFs of two groups were found (Figure 3). During running, elderly loaded their hip joints ~ 2.5 BW less than the young subjects for all respective speeds. Also, hip joint loading did not increase in elderly between walking and running at similar speeds whereas in the young subjects hip joint loading increased by ~ 2 BW. Lower hip adduction moments during running in elderly compared to young (Supplemental Figure 1) and the similar hip adduction moments during walking and running at similar speeds in elderly (Supplemental Figure 1) explain the absence of increased hip loading in running. Two mechanisms contribute to this: a) vertical GRF magnitude is decreased in higher running speeds, by decreasing peak vertical acceleration through reducing step length [22]; b) the frontal plane lever arm of the GRF with respect to the hip joint is decreased by decreasing their hip adduction angle during peak loading. Whether this adjustment relates to neuromuscular decline due to aging or other compensatory protective mechanisms is yet to be found.

Limiting assumptions intrinsic to the musculoskeletal modeling [18], must be taken into consideration. Muscle contraction dynamics are not taken into account in static optimization, thus muscle force levels are calculated to instantaneously satisfy the moment equilibrium and may not be representative of real muscle activations. The generic musculoskeletal model used in this study does not consider the subject-specific effect of aging on muscle physiology. Moreover, documented differences in kinematics and kinetics between overground and treadmill conditions during walking and running in healthy individuals [23,24] suggest caution when extrapolating our findings to overground walking/running. However, those differences were “typically less than or close to the variation present in the measures

1 themselves”[24]. In addition, small sample size in some speeds is a limiting factor in the
2 statistical comparisons and regression analysis.

3 With this work, hip joint loading is reported in healthy elderly women during walking
4 and running at different speeds, and related to specific adaptation in motion strategy and
5 compared to a young population. The elderly group showed less stride length and hip
6 adduction angle at peak loading during both walking and running. That strategy resulted in
7 reduced peak HCFs in elderly only during running and similar peak HCFs during walking and
8 running at the same speeds in elderly. That suggested that running may not have an added
9 value compared to walking in terms of peak hip joint loading in elderly. These conclusions
10 must be considered when designing specific training programs for young or elderly adults that
11 aim at preserving or enhancing hip bone mass.

13 **Conflict of Interest statement**

14 All authors state that they have no conflicts of interest.

Figure legends

Figure 1 Ensemble curves of HCF in different speeds during walking and running along the stance phase for both elderly and young adults. Positive HCFs indicate downward direction.

Figure 2 Ensemble curves of GRF in different speeds during walking and running along the stance phase for both elderly and young adults. Positive GRFs indicate upward direction.

Figure 3 Peak HCFs and GRFs for both elderly and young groups in every speed during walking. Asterisks denote significant ($p < 0.05$) comparisons between groups per speed.

Supplemental Figure 1: Ensemble curves of hip moments at three planes of motion for elderly (A) and young (B) adults.

Supplemental Table 1: Mean (SD) peak HCFs and GRFs during walking and running for both elderly and young groups.

Figures

1 **Figure 1**

2

3

Figure 2

4

5

6

7

8

9

Figure 3

References

- [1] E. Hernlund, a Svedbom, M. Ivergård, J. Compston, C. Cooper, J. Stenmark, E. V McCloskey, B. Jönsson, J. a Kanis, Osteoporosis in the European Union: medical management, epidemiology and economic burden. A report prepared in collaboration with the International Osteoporosis Foundation (IOF) and the European Federation of Pharmaceutical Industry Associations (EFPIA)., *Arch. Osteoporos.* 8 (2013) 136. doi:10.1007/s11657-013-0136-1.
- [2] R. Marcus, Role of exercise in preventing and treating osteoporosis, *Rheum. Dis. Clin. North Am.* 27 (2001) 131–141.
- [3] K.T. Borer, Physical activity in the prevention and amelioration of osteoporosis in women : interaction of mechanical, hormonal and dietary factors., *Sports Med.* 35 (2005) 779–830.
- [4] D. Kerr, a Morton, I. Dick, R. Prince, Exercise effects on bone mass in postmenopausal women are site-specific and load-dependent., *J. Bone Miner. Res.* 11 (1996) 218–25. doi:10.1002/jbmr.5650110211.
- [5] M. Martyn-St James, S. Carroll, A meta-analysis of impact exercise on postmenopausal bone loss: the case for mixed loading exercise programmes., *Br. J. Sports Med.* 43 (2009) 898–908. doi:10.1136/bjsm.2008.052704.
- [6] M. Martyn-St James, S. Carroll, Meta-analysis of walking for preservation of bone mineral density in postmenopausal women., *Bone.* 43 (2008) 521–31. doi:10.1016/j.bone.2008.05.012.
- [7] B. a Wallace, R.G. Cumming, Systematic review of randomized trials of the effect of exercise on bone mass in pre- and postmenopausal women., *Calcif. Tissue Int.* 67

(2000) 10–18. doi:10.1007/s00223001089.

[8] T.E. Howe, B. Shea, L.J. Dawson, F. Downie, A. Murray, C. Ross, R.T. Harbour, L.M. Caldwell, G. Creed, Exercise for preventing and treating osteoporosis in postmenopausal women, *Cochrane Libr.* (2011).

[9] P.J. Ehrlich, L.E. Lanyon, International Review Article Mechanical Strain and Bone Cell Function : A Review, (2002) 688–700.

[10] H.M. Frost, Bone's mechanostat: a 2003 update, *Anat. Rec. Part A Discov. Mol. Cell. Evol. Biol.* 275 (2003) 1081–1101.

[11] E.J. Bassey, S.J. Ramsdale, Weight-bearing exercise and ground reaction forces: a 12-month randomized controlled trial of effects on bone mineral density in healthy postmenopausal women., *Bone*. 16 (1995) 469–476.

[12] K.T. Borer, K. Fogleman, M. Gross, J.M. La New, D. Dengel, Walking intensity for postmenopausal bone mineral preservation and accrual., *Bone*. 41 (2007) 713–721. doi:10.1016/j.bone.2007.06.009.

[13] G. Giarmatzis, I. Jonkers, M. Wesseling, S. Van Rossom, S. Verschueren, Loading of hip measured by hip contact forces at different speeds of walking and running, *J. Bone Miner. Res.* 30 (2015) 1431–1440.

[14] P. DeVita, T. Hortobagyi, Age causes a redistribution of joint torques and powers during gait, *J. Appl. Physiol.* 88 (2000) 1804–1811.

[15] R.K. Fukuchi, M. Duarte, Comparison of three-dimensional lower extremity running kinematics of young adult and elderly runners., *J. Sports Sci.* 26 (2008) 1447–1454. doi:10.1080/02640410802209018.

[16] S.R. Hamner, A. Seth, S.L. Delp, Muscle contributions to propulsion and support

during running., J. Biomech. 43 (2010) 2709–16. doi:10.1016/j.jbiomech.2010.06.025.

[17] F. DeGroot, T. De Laet, I. Jonkers, J. De Schutter, Kalman smoothing improves the estimation of joint kinematics and kinetics in marker-based human gait analysis, J. Biomech. 41 (2008) 3390–3398.

[18] A. Erdemir, S. McLean, W. Herzog, A.J. van den Bogert, Model-based estimation of muscle forces exerted during movements., Clin. Biomech. (Bristol, Avon). 22 (2007) 131–154. doi:10.1016/j.clinbiomech.2006.09.005.

[19] G. Lenaerts, M. Mulier, a Spaepen, G. Van der Perre, I. Jonkers, Aberrant pelvis and hip kinematics impair hip loading before and after total hip replacement., Gait Posture. 30 (2009) 296–302. doi:10.1016/j.gaitpost.2009.05.016.

[20] G. Bergmann, G. Deuretzbacher, M. Heller, F. Graichen, A. Rohlmann, J. Strauss, G.N. Duda, Hip contact forces and gait patterns from routine activities, J. Biomech. 34 (2001) 859–871.

[21] G. Bergmann, F. Graichen, A. Rohlmann, Hip joint loading during walking and running, measured in two patients, J. Biomech. 26 (1993) 969–990.

[22] B.C. Heiderscheit, E.S. Chumanov, M.P. Michalski, C.M. Wille, M.B. Ryan, Effects of step rate manipulation on joint mechanics during running, Med. Sci. Sports Exerc. 43 (2011) 296.

[23] S.J. Lee, J. Hidler, Biomechanics of overground vs. treadmill walking in healthy individuals., J. Appl. Physiol. 104 (2008) 747–55. doi:10.1152/jappphysiol.01380.2006.

[24] J.R. Watt, J.R. Franz, K. Jackson, J. Dicharry, P.O. Riley, D.C. Kerrigan, A three-dimensional kinematic and kinetic comparison of overground and treadmill walking in

healthy elderly subjects, Clin. Biomech. 25 (2010) 444–449.
doi:10.1016/j.clinbiomech.2009.09.002.

[2] R. Marcus, Role of exercise in preventing and treating osteoporosis, Rheum. Dis. Clin. North Am. 27 (2001) 131–141.

[3] K.T. Borer, Physical activity in the prevention and amelioration of osteoporosis in women : interaction of mechanical, hormonal and dietary factors., Sports Med. 35 (2005) 779–830.

[4] D. Kerr, a Morton, I. Dick, R. Prince, Exercise effects on bone mass in postmenopausal women are site-specific and load-dependent., J. Bone Miner. Res. 11 (1996) 218–25.

[5] M. Martyn-St James, S. Carroll, A meta-analysis of impact exercise on postmenopausal bone loss: the case for mixed loading exercise programmes., Br. J. Sports Med. 43 (2009) 898–908.

[6] M. Martyn-St James, S. Carroll, Meta-analysis of walking for preservation of bone mineral density in postmenopausal women., Bone. 43 (2008) 521–31.

[7] B. a Wallace, R.G. Cumming, Systematic review of randomized trials of the effect of exercise on bone mass in pre- and postmenopausal women., Calcif. Tissue Int. 67 (2000) 10–18.

[8] T.E. Howe, B. Shea, L.J. Dawson, F. Downie, A. Murray, C. Ross, R.T. Harbour, L.M. Caldwell, G. Creed, Exercise for preventing and treating osteoporosis in postmenopausal women, Cochrane Libr. (2011).

[9] P.J. Ehrlich, L.E. Lanyon, International Review Article Mechanical Strain and Bone

Cell Function : A Review, (2002) 688–700.

[10] H.M. Frost, Bone's mechanostat: a 2003 update, *Anat. Rec. Part A Discov. Mol. Cell. Evol. Biol.* 275 (2003) 1081–1101.

[11] E.J. Bassey, S.J. Ramsdale, Weight-bearing exercise and ground reaction forces: a 12-month randomized controlled trial of effects on bone mineral density in healthy postmenopausal women., *Bone*. 16 (1995) 469–476.

[12] K.T. Borer, K. Fogleman, M. Gross, J.M. La New, D. Dengel, Walking intensity for postmenopausal bone mineral preservation and accrual., *Bone*. 41 (2007) 713–721.

[13] G. Giarmatzis, I. Jonkers, M. Wesseling, S. Van Rossom, S. Verschueren, Loading of hip measured by hip contact forces at different speeds of walking and running, *J. Bone Miner. Res.* 30 (2015) 1431–1440.

[14] P. DeVita, T. Hortobagyi, Age causes a redistribution of joint torques and powers during gait, *J. Appl. Physiol.* 88 (2000) 1804–1811.

[15] R.K. Fukuchi, M. Duarte, Comparison of three-dimensional lower extremity running kinematics of young adult and elderly runners., *J. Sports Sci.* 26 (2008) 1447–1454.

[16] S.R. Hamner, A. Seth, S.L. Delp, Muscle contributions to propulsion and support during running., *J. Biomech.* 43 (2010) 2709–16.

[17] F. DeGroote, T. De Laet, I. Jonkers, J. De Schutter, Kalman smoothing improves the estimation of joint kinematics and kinetics in marker-based human gait analysis, *J. Biomech.* 41 (2008) 3390–3398.

[18] A. Erdemir, S. McLean, W. Herzog, A.J. van den Bogert, Model-based estimation of muscle forces exerted during movements., *Clin. Biomech. (Bristol, Avon)*. 22 (2007) 131–154.

- [19] G. Lenaerts, M. Mulier, a Spaepen, G. Van der Perre, I. Jonkers, Aberrant pelvis and hip kinematics impair hip loading before and after total hip replacement., *Gait Posture*. 30 (2009) 296–302.
- [20] G. Bergmann, G. Deuretzbacher, M. Heller, F. Graichen, A. Rohlmann, J. Strauss, G.N. Duda, Hip contact forces and gait patterns from routine activities, *J. Biomech.* 34 (2001) 859–871.
- [21] G. Bergmann, F. Graichen, A. Rohlmann, Hip joint loading during walking and running, measured in two patients, *J. Biomech.* 26 (1993) 969–990.
- [22] B.C. Heiderscheit, E.S. Chumanov, M.P. Michalski, C.M. Wille, M.B. Ryan, Effects of step rate manipulation on joint mechanics during running, *Med. Sci. Sports Exerc.* 43 (2011) 296.
- [23] S.J. Lee, J. Hidler, Biomechanics of overground vs. treadmill walking in healthy individuals., *J. Appl. Physiol.* 104 (2008) 747–55.
- [24] a G. Schache, P.D. Blanch, D. a Rath, T. V Wrigley, R. Starr, K.L. Bennell, A comparison of overground and treadmill running for measuring the three-dimensional kinematics of the lumbo-pelvic-hip complex., *Clin. Biomech. (Bristol, Avon)*. 16 (2001) 667–680.