www.mednet.gr/archives
ARCHIVES OF HELLENIC MEDICINE: ISSN 11-05-3992

ORIGINAL PAPER
ΕΡΕΥΝΗΤΙΚΗ ΕΡΓΑΣΙΑ

ARCHIVES OF HELLENIC MEDICINE 2025, 42(3):390–396
ΑΡΧΕΙΑ ΕΛΛΗΝΙΚΗΣ ΙΑΤΡΙΚΗΣ 2025, 42(3):390–396

The effects of abdominal obesity on the heel strike transient during walking by using simulated abdominal weights on healthy people A pilot study

E. Paraskevopoulos,[1,2]
T. Tsatalas,[3]
R. Levantinou,[3]
G. Giakas[3]

[1]Department of Physiotherapy, University of West Attica, Athens
[2]Department of Physiotherapy, Aegean College, Athens
[3]Department of Physical Education and Sport Science, University of Thessaly, Trikala, Greece

OBJECTIVE **To identify whether abdominal weights increase heel strike transients (HST) in healthy individuals.** METHOD **Ten experimental subjects (mean age 27.5±8.84) with external weights placed on their abdomen, under specific experimental conditions, walked over a 9.5 walkway instrumented with a Bertec force plate at a self-selected low and high speed. Friedman's ANOVA test was used to detect any significant differences between conditions on the HST, the first and second peak maximum, the contact time and the loading rate.** RESULTS **The HST changed significantly only when the subjects increased their gait speed. However, the first maximum force peak increased significantly when the external weights were placed on them, under specific experimental conditions. The second maximum force peak decreased significantly when the external weights were placed on the subjects, under specific experimental conditions.** CONCLUSIONS **Abdominal weight may not affect the size of the HST and rates of loading in healthy individuals. However, the first maximum force may increase on healthy participants.**

Οι επιδράσεις της κοιλιακής παχυσαρκίας στην πρώτη παροδική κατακόρυφη εδαφική δύναμη αντίδρασης κατά τη βάδιση χρησιμοποιώντας προσομοιωμένα κοιλιακά βάρη σε υγιή άτομα: Μια πιλοτική μελέτη

Περίληψη στο τέλος του άρθρου

Key words

Cartilage
Force plate
Gait analysis
Knee
Osteoarthritis

Submitted 24.3.2024
Accepted 27.4.2024

The ground reaction force (GRF) serves as a valuable marker in the field of biomechanics, facilitating the assessment of musculoskeletal injury risk. GRF patterns play a crucial role in biomechanical analyses, aiding in the evaluation of both normal and pathological gait.[1] Moreover, the repeated occurrence of unusually high GRFs has been linked to an increased risk of injury and the development of osteoarthritis (OA).[2,3] Research indicates that postmenopausal women tend to exhibit higher levels of abdominal fat mass compared to premenopausal women.[4,5]

Estrogens are believed to influence the distribution of body fat in women, particularly leading to a higher concentration of abdominal fat. Previous research studies suggest that administering exogenous estrogen to postmenopausal women may lead to lower waist-to-hip ratios and reduced visceral adipose tissue compared to women without estrogen supplementation.[5] Furthermore, obesity has been identified as a significant risk factor for knee OA. Studies have shown that obesity increases the loading on the knee joint, triggering the activation of chondrocyte mechanoreceptors. This activation, in turn, leads to the release of cytokines, growth factors, and metalloproteinases, which interfere with matrix synthesis and contribute to cartilage degeneration and the development of OA.[3]

Research has demonstrated that repeated uncontrollable impulse loading of the bone increases the risk of microfractures, leading to stiffening during the healing process. Additionally, this repetitive loading pattern decreases the duration of eccentric activation of the quadriceps, resulting in rapid muscle fatigue and alterations in ground reaction forces.[6] This mechanism increases the rate of loading on the knee joint because the musculature, particularly the quadriceps muscle, serves as a shock-absorbing mechanism.[7–9]

At heel strike, vertical impact forces initiate transient stress waves that propagate through the lower extremities, traveling up the kinetic chain.[10] These transient stress waves are called heel strike transients (HST).[11] HST's are generated from the sudden impact of the heel with the ground surface. The magnitude of these ground reaction forces is largely depending on gait speed, footwear, hardness of the ground surface and the angle and velocity of the foot.[11–13] HST's may appear in approximately one third of the general population during walking and the reason for their existence in some people is still unclear.[14] However, previous suggestions indicate that the HST is a consequence of increased rates of loading. Furthermore, research supports a positive association between HST and the etiology, as well as the progression of OA.[10,14]

To date, the majority of studies have aimed to identify possible correlations between the presence of the HST and muscular weakness of the lower limb, as well as the development and presence of OA.[11,15] However, conflicting evidence exists in the literature regarding the aforementioned statements.[14,16–18] It is widely acknowledged that there is a relationship between hormonal changes occurring during menopause and the development of knee OA.[19–22] Given the alteration of fat distribution in postmenopausal women and the associated increase in the prevalence and incidence of knee OA,[23] it is crucial to identify potential mechanisms that may influence joint integrity. Therefore, the purpose of this study was to investigate the impact of different levels of external weight on GRF and the HST during walking at freely chosen high and low speeds.

MATERIAL AND METHOD

The study protocol received approval from the Ethics Committee of the Department of Physical Education and Sports Science at the University of Thessaly in Trikala. All procedures were reviewed and approved by the Internal Ethics Committee of the Department on April 23, 2015 (protocol number: 1000).

Participants

Due to the weight carriage component of the study, several exclusion criteria were applied. Participants were excluded if they had a history of lower limb or spinal injury or surgical treatment, painful knees, or developmental or congenital abnormalities of the lower limbs. Additionally, individuals with neurological conditions affecting the lower limbs, such as paralysis or paresis, or other medical problems altering normal gait patterns were excluded. Participants were also ineligible if they had a history of cerebrovascular disease, hernia, rheumatoid arthritis, autoimmune diseases affecting gait patterns, spinal stenosis, or chronic or acute back pain. The exclusion criteria were explained to potential participants, and those meeting the criteria were asked to sign a consent form before participation. Finally, ten subjects (seven males and three females) were recruited for this study. The subject's age, height and weight were (mean±standard deviation) 27.5±8.84, 174.6±3.83 and 71.5±6.36, respectively. Their average body mass index (BMI) and waist to hip ratio were 23.5±1.7 and 0.79±0.05, respectively.

Equipment

In the lab, a Bertec force plate was employed to measure the GRF. This force plate is specifically designed to measure forces in three perpendicular axes and moments and their associated axes. The output data were recorded in a text file containing all the necessary measurements, along with a title for easy identification of the data collected.[24,25] The biomechanical gait analysis was conducted on a 9.5 meters level walkway equipped with a Bertec force plate. This setup allowed for the measurement of both the magnitude and direction of the GRF applied by the foot to the ground.[10] The HST can be observed by utilizing force plates at the onset of the stance phase during gait analysis. However, it has been overlooked by some investigators who have utilized force plates due to limitations such as insufficient high-frequency sampling rates and excessive low-pass filtering.[10]

Procedure

The testing equipment was calibrated prior to commencement of testing. Weight was measured on the force platform in Newtons, and height was measured using a stadiometer. Each participant walked over the force platform with their dominant foot at both a self-selected low and high speed, repeating the process three times. Only trials where the entire foot landed on the force platform and the gait pattern remained unaltered were accepted. Participants were instructed to walk freely along the walkway without specifically targeting the force plate. Since visual guidance does not influence GRF variability,[26,27] it was not considered essential for the subjects to wear special goggles in order to reduce peripheral vision and eliminate force plate targeting as previously described.[28]

The subjects were instructed to maintain forward gaze at eye level and to refrain from looking directly at the force plates during the trials. Participants, being healthy and sufficiently fit, practiced their walking trials multiple times to familiarize themselves with the dimensions of the walkway and the positioning of the force plates. Initial trials were conducted to determine the optimal starting point for each subject and to ensure that their dominant leg would strike the force plate. These preliminary trials were performed three times without any external weight.

Subsequently, a custom-made device resembling a backpack was positioned on the side of the belly, with ample space to accommodate 5 kg, 10 kg, and 15 kg weight plates affixed to the front of the abdomen. Since the additional weight was positioned on the anterior-external part of the abdomen, it was considered to simulate subcutaneous fat, which is the fat located closest to the skin on the outermost part of the abdomen. This positioning was chosen to closely replicate the effects of external weight. The subjects were then instructed to perform three trials with each of the 5 kg, 10 kg, and 15 kg external weights.

The protocol was conducted at both a low walking speed, representing their typical relaxed daily walking pace, and at a higher speed, simulating rushed walking. Prior to the experimental trials, participants engaged in practice sessions to ensure that their foot would consistently strike the force plate when the external weights were applied, and to determine the appropriate starting point for each trial.[26] The duration taken to reach the force plate without external weight and with each external weight was carefully monitored using a timer to detect any changes in gait velocity. An independent observer was consistently present to ensure that measurements were conducted under consistent environmental conditions.[26]

Data processing

To ensure data integrity, the observer verified that the investigator collected force plate data at a sampling rate of 1,000 Hz. Force plates were manually activated after instructing each subject to begin their walking trial from a predetermined starting point. The software provided various data, which were transferred to Microsoft Excel 2007. For each valid trial, the program generated a force curve graph.

GRF data collected from subjects were normalized to body weight (BW) for trials without external weights. For trials with external weights, data were normalized to BW plus the weight of the external load. Each trial was labelled with the subject's identification number and indicated the presence and magnitude of the external weight.

Mean readings were taken for several parameters, including the heel-strike transient (HST), the first maximum force peak (FMFP), the second maximum force peak (SMFP), the time taken for the FMFP to occur, and the total contact time, as described previously (fig. 1).[26] Also, the time to first peak along with the loading rate of the first peak maximum were calculated by dividing the FMFP by the time to FMFP.[26,29]

The aim of this study was to identify statistically significant differences in the GRF components, total contact time, and loading rates of the HST and the FMFP across various alterations of external weight and different walking velocities. A total of 240 measurements were analysed. Data were exported to Microsoft Excel 2007 to derive the required dependent variables. A graph was generated for each trial, and values were extracted from these graphs. The mean value of the three accepted trials was calculated and compared using the appropriate statistical tests in the Statistical Package for Social Sciences (SPSS), version 18.0 for Windows.

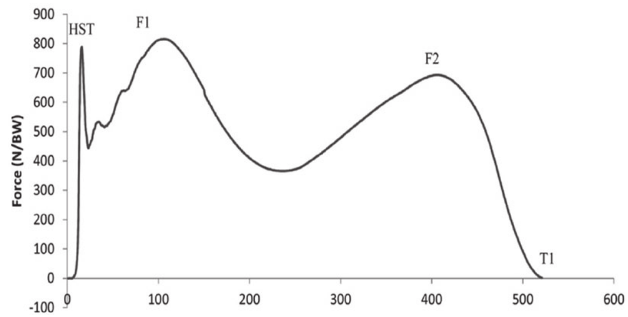

Figure 1. Force/time curve demonstrating the heel strike transient (HST), the first maximum force peak (F1), the second maximum force peak (F2) and the total contact time (T1). BW: Body weight.

The dependent variables were calculated and defined as follows: The HST was defined as the vertical force recorded prior to the usual, expected peak vertical force derived from heel strike, calculated via the force plate. A specific way to identify HSTs is not available in the literature and most researchers use their own rationale to measure HSTs.[17] The maximal vertical force recorded after the obvious HST (if the HST is present), calculated via the force plate. The total contact time as it was evident by the graph, calculated via the force plate. The loading rates were calculated by dividing the maximal vertical force by the time to the maximal vertical force.[30]

Statistical analysis

To determine the appropriate test for comparing the various trials, the normality of the data was assessed using the Kolmogorov-Smirnov test. Additionally, means and standard deviations (SD) were calculated to provide descriptive statistics for the data. The One-way repeated measures analysis of variance (ANOVA) test was found to be the appropriate test for each dependent variable. This test was used to discover statistically significant differences between the external weights and then between the different velocities for the HST, the FMFP, the total contact time and the loading rates. Differences were considered significant at the 5% level. Data analyses were conducted using the Statistical Package for the Social Sciences (SPSS), version 25.0 (SPSS Inc, Chicago, IL, USA).

RESULTS

Based on the findings of the normality examination it was evident that the data were not normally distributed. Since the data did not appear to be normally distributed, a nonparametric test was used. Friedman's ANOVA is the nonparametric alternative to the One-way ANOVA with repeated measures. Friedman's ANOVA was used to evaluate differences across several different measurement conditions for each variable. The Friedman's ANOVA test showed a statistically significant difference at the HST in the eight different conditions measured ($p<0.001$). Post-hoc analysis with Wilcoxon signed-rank test was conducted for all the variables tested with a Bonferroni correction applied, resulting in a significance level set at $p<0.00625$. The results are available in table 1.

The mean and SD analysis indicates a predominant increase in the HST as gait velocity increased. The statistically significant difference ($p<0.001$) observed can be attributed to variations in gait velocity. Similarly, for the FMFP, there was a gradual increase with both increasing external weight and gait velocity. The statistical significance ($p<0.001$) observed was evident across multiple comparisons. Regarding the SMFP, there were significant differences ($p<0.001$) in several comparisons, yet these differences were not attributed to variations in gait velocity. Interestingly, while the HST and FMFP increased with added weight and increased gait velocity, the SMFP showed a slight decrease in most subjects under these conditions. Overall, these findings suggest that adding external weight and increasing gait velocity have differential effects on various components of the ground reaction force, with the SMFP showing a unique response compared to the HST and FMFP.

The time taken to reach the first maximum vertical force peak was significantly reduced only when subjects increased their gait velocity. The observed statistical significance ($p<0.001$) in the mean time to reach the first peak was primarily attributed to the increase in gait velocity. Similarly, the contact time of the foot with the force plate increased significantly ($p<0.001$) only when subjects altered their gait velocity. Total contact time serves as a reliable indicator of changes in gait velocity.[26] Interestingly, total contact time did not significantly change when external weights were added to the subjects. The loading rate significantly changed with variations in gait velocity ($p<0.001$). Notably, the loading rate was greater when subjects walked on the force plate without any external weights in both slow and fast gait conditions. Overall, these results underscore the influence of gait velocity on various parameters such as time to reach the first peak, contact time, and loading rate, with minimal impact observed from the addition of external weights.

DISCUSSION

The findings of this study suggest that the size of the HST is not influenced by the addition of external weight simulating the accumulation of abdominal fat. The study predominantly involved young and active subjects, which supports the notion that the muscles of the thigh serve as the primary shock absorbers during walking. It has been previously supported that quadriceps strength produces a breaking action that is essential for knee deceleration just before heel-strike.[31] Therefore, quadriceps weakness could potentially lead to an increase in the size of the HST and consequently increase the impact loading of the knee. However, the ability of the subjects to practice their gait technique with external weights allowed them to adequately prepare their muscles, enabling them to activate the necessary reflexes and absorb the energy of impact by appropriately lengthening the muscles around the knee joint.[32]

Although the HST and the loading rates were not significantly different when the external weights were placed on the subjects, the size of the FMFP was significantly increased when external weights were placed on the subjects. This finding aligns with a previous study[33] which demonstrated that the absolute peak vertical GRF increases proportionally with BW. However, it is important to note that the previous study recruited individuals diagnosed with knee OA,

Table 1. Results from force plate measurements.

	Mean HST (N/BW)	1st peak max (N/BW)	2nd peak max (N/BW)	Time to 1st peak (sec)	Contact time (millisec)	Loading rate (BW/sec)
Slow 0 kg	0.79	1.19	1.15	0.12	624.10	13.40
Slow 5 kg	0.78	1.24	1.15	0.14	625.20	8.78
Slow 10 kg	0.79	1.27	1.11	0.13	633.80	9.64
Slow 15 kg	0.71	1.30	1.06	0.14	612.65	9.78
Fast 0 kg	1.08	1.39	1.18	0.10	513.10	16.12
Fast 5 kg	1.09	1.47	1.12	0.11	513.70	13.96
Fast 10 kg	1.06	1.50	1.05	0.11	513.90	14.60
Fast 15 kg	1.08	1.51	99.00	0.09	506.10	16.68

BW: Body weight, Slow: Low gait speed, Fast: Fast gait speed, Mean HST: Mean heel strike transient, 1st peak max: 1st vertical force peak maximum; 2nd peak max: 2nd vertical force peak maximum, Contact time: Contact time on force plate, Loading rate: Representation of how quickly the impact force is applied and therefore, it did not provide information about GRF in obese individuals without knee OA. Additionally, research indicates that statistical significance was only observed in the absolute GRFs.[33] In this current study the statistical significance was evident even after normalizing the GRFs for BW. Previous research has shown that the size of the GRF may be influenced by walking speed.[34] However, in this study, the gait speed did not change when external weights were placed on the subjects, as indicated by the statistically unchanged contact time.[35,36] Additionally, the size of the HST did not significantly change at any point with the placement of external weights, an aspect strongly influenced by gait speed.[13]

It should be especially noted that significant alterations in gait velocity would have created statistically significant differences in foot contact time.[35] Moreover, previous research studies have also reported that obese individuals exhibit larger vertical GRF when compared with normal-weight individuals.[34] However, in their study the comparison was made with absolute and not with normalized GRFs. Still, the statistical significance in the current study was evident with and without normalizing the GRFs.

Interestingly, the SMFP, representing the vertical GRF during the propulsive phase of gait, exhibited a decrease when external weights were added, and this decrease was further pronounced with an increase in gait velocity, contrasting with the behaviour of the FMFP. In some instances, there was a statistically significant difference in the SMFP between conditions, regardless of gait speed. In this study, the addition of external weights led to the development of larger GRFs during the passive part of gait, which may have detrimental implications for the integrity of the knee joint.

Research evidence has shown that heel contact produces larger forces that are easily transmitted to the knee joint.[37] In contrast, landing with the metatarsal heads may attenuate the GRFs that may reach the knee joint and reduce the risk of knee degeneration. The loading rates increased significantly only when gait velocity increased. It seems that the absence of knee OA and similarly quadriceps muscle weakness may not affect the loading rates of the lower limbs when adding external weights. This finding is also supported by more researchers who state that people with knee OA and muscle weakness tend to have higher loading rates.[38]

Furthermore, an interesting study showed that people with pre-osteoarthritic changes may have distinct HSTs and as a result higher loading rates.[39] Furthermore, the study revealed that the pre-osteoarthritic group exhibited shorter periods of eccentric contraction of the quadriceps as detected by electromyography (EMG). This finding reinforces the idea that the quadriceps muscle possesses shock-absorbing properties. Notably, avoidance of quadriceps engagement during ambulation is frequently observed in individuals with knee OA, particularly in those who are obese.[40] This gait pattern transfers significant amount of impact shock to the knee joint which reduces the thickness of the cartilage and joint space. This mechanism also makes the ligaments of the knee less stable.[14] Also, their ligamentous laxity may be reduced only by the muscles of the thigh and if their weak their loading rates may increase by 21% when compared with people with stronger muscles.[41] It should be noted that higher loading rates may be seen irrespective of BW or knee joint integrity in people with weak lower limb muscles.[14]

Limitations

A limitation of this study is that the investigator is not aware of any possible altered rear-foot motions that may have been used by the subjects in order to reduce the vertical impact of their lower limbs. The everted rear foot posture in their experimental group reduced the impact force during ambulation and resulted in a smaller and delayed HST.[11] Since the first and second peak maximum were statistically different in multiple weight conditions but with the same speed, it is unknown whether the subjects used different multi-joint compensatory strategies in order to maintain the size of the HST unchanged when the external weights were added.[42–45]

Furthermore, another limitation is the recruitment of a gender-mixed group of people. The inclusion of a female group of people would have been more appropriate for this study. However, this study did not compare the findings between the participants. This study assessed the same people before and after the intervention and thus measuring different genders should not be a problem for the identification of HST alterations when external weights were added.

In conclusion, the findings of this study suggest that increased abdominal fat may indeed alter the gait pattern. However, it remains unknown whether compensatory strategies involving the foot or the upper joints of the lower limbs and spine may influence the size of the vertical GRF. The study underscores the critical role of quadriceps strength and gait technique in knee loading during walking. External weight addition didn't significantly alter knee loading, suggesting factors beyond weight influence impact. Despite normalization, increased GRFs indicate multifactorial knee loading beyond obesity. Unique gait patterns observed in knee OA individuals emphasize the importance of tailored interventions. Gender considerations warrant attention for comprehensive understanding. Overall, interventions targeting muscle strength, gait mechanics, and weight management could mitigate knee loading, reducing injury risks and enhancing knee health in clinical and public health settings.

ACKNOWLEDGEMENTS

The authors would like to thank the University of Thessaly and the members of the biomechanics lab for their assistance. Most of all, we would like to thank the participants of the study, without whom this work would not have been possible.

ΠΕΡΙΛΗΨΗ

Οι επιδράσεις της κοιλιακής παχυσαρκίας στην πρώτη παροδική κατακόρυφη εδαφική δύναμη αντίδρασης κατά τη βάδιση χρησιμοποιώντας προσομοιωμένα κοιλιακά βάρη σε υγιή άτομα: Μια πιλοτική μελέτη

Ε. ΠΑΡΑΣΚΕΥΟΠΟΥΛΟΣ,[1,2] Θ. ΤΣΑΤΑΛΑΣ,[3] Ρ. ΛΕΒΑΝΤΙΝΟΥ,[3] Γ. ΓΙΑΚΑΣ[3]

[1]Τμήμα Φυσικοθεραπείας, Πανεπιστήμιο Δυτικής Αττικής, Αθήνα, [2]Τμήμα Φυσικοθεραπείας, Aegean College, Αθήνα, [3]Τμήμα Επιστήμης Φυσικής Αγωγής και Αθλητισμού, Πανεπιστήμιο Θεσσαλίας, Τρίκαλα

Αρχεία Ελληνικής Ιατρικής 2025, 42(3):390–396

ΣΚΟΠΟΣ Να εξετάσει εάν τα κοιλιακά βάρη αυξάνουν το heel strike transient (HST) σε υγιή άτομα. **ΥΛΙΚΟ-ΜΕΘΟΔΟΣ** Προσήλθαν 10 άτομα από δειγματοληψία ευκολίας, τα οποία εξετάστηκαν ως προς την καταλληλότητά τους για τη μελέτη. Τα άτομα αυτά βάδισαν σε έναν διάδρομο 9,5 m, στον οποίο είχε τοποθετηθεί δυναμοδάπεδο Bertec. Από τους εξεταζόμενους ζητήθηκε να περάσουν από το δυναμοδάπεδο με το κυρίαρχο κάτω άκρο σε μια ορισμένη χαμηλή και υψηλή ταχύτητα, όπως αυτή επιλέχθηκε από τους εξεταζόμενους. Η δοκιμασία Friedman's ANOVA χρησιμοποιήθηκε για να εντοπίσει στατιστικά σημαντικές διαφορές μεταξύ των συνθηκών. **ΑΠΟΤΕΛΕΣΜΑΤΑ** Τα αποτελέσματα έδειξαν ότι το HST, ο χρόνος μέχρι την πρώτη μέγιστη κατακόρυφη δύναμη, ο χρόνος επαφής και ο ρυθμός φόρτισης επηρεάστηκαν μόνο από την ταχύτητα βάδισης και όχι από την τοποθέτηση βαρών. Παρ' όλα αυτά, η πρώτη μέγιστη κατακόρυφη δύναμη αυξήθηκε στατιστικώς σημαντικά με την αλλαγή της ταχύτητας βάδισης αλλά και με την αύξηση των εξωτερικών βαρών και, αντίθετα, μειώθηκε η δεύτερη μέγιστη κατακόρυφη δύναμη. **ΣΥΜΠΕΡΑΣΜΑΤΑ** Το κοιλιακό βάρος μπορεί να μην επηρεάζει το μέγεθος του HST και τα ποσοστά φόρτισης σε υγιή άτομα. Ωστόσο, η πρώτη μέγιστη δύναμη ενδέχεται να αυξηθεί στους υγιείς συμμετέχοντες.

Λέξεις ευρετηρίου: Ανάλυση βάδισης, Γόνατο, Δυναμοδάπεδο, Οστεοαρθρίτιδα, Χόνδρος

References

1. MUNRO CF, MILLER DI, FUGLEVAND AJ. Ground reaction forces in running: A reexamination. *J Biomech* 1987, 20:147–155
2. CAVANAGH PR, LAFORTUNE MA. Ground reaction forces in distance running. *J Biomech* 1980, 13:397–406
3. BRAY G, BOUCHARD C. *Handbook of obesity: Clinical applications.* 4th ed. CRC Press, London, 2014
4. OCHI M, TABARA Y, KIDO T, UETANI E, OCHI N, IGASE M ET AL. Quadriceps sarcopenia and visceral obesity are risk factors for postural instability in the middle-aged to elderly population. *Geriatr Gerontol Int* 2010, 10:233–243
5. LOVEJOY JC, CHAMPAGNE CM, DE JONGE L, XIE H, SMITH SR. Increased visceral fat and decreased energy expenditure during the menopausal transition. *Int J Obes (Lond)* 2008, 32:949–958
6. MESSIER SP, DAVIES AB, MOORE DT, DAVIS SE, PACK RJ, KAZMA SC. Severe obesity: Effects on foot mechanics during walking. *Foot Ankle Int* 1994, 15:29–34
7. HILLS AP, HENNIG EM, BYRNE NM, STEELE JR. The biomechanics of adiposity – structural and functional limitations of obesity and implications for movement. *Obes Rev* 2002, 3:35–43
8. SYED IY, DAVIS BL. Obesity and osteoarthritis of the knee: Hypotheses concerning the relationship between ground reaction forces and quadriceps fatigue in long-duration walking. *Med Hypotheses* 2000, 54:182–185
9. MESSIER SP. Osteoarthritis of the knee and associated factors of age and obesity: Effects on gait. *Med Sci Sports Exerc* 1994, 26:1446–1452
10. COLLINS JJ, WHITTLE MW. Impulsive forces during walking and their clinical implications. *Clin Biomech (Bristol, Avon)* 1989, 4:179–187
11. LEVINGER P, GILLEARD W. The heel strike transient during walking in subjects with patellofemoral pain syndrome. *Phys Ther Sport* 2005, 6:83–88
12. LAFORTUNE MA, HENNIG EM. Cushioning properties of footwear during walking: Accelerometer and force platform measure-

ments. *Clin Biomech (Bristol, Avon)* 1992, 7:181–184
13. WHITTLE MW. Generation and attenuation of transient impulsive forces beneath the foot: A review. *Gait Posture* 1999, 10:264–275
14. MIKESKY AE, MEYER A, THOMPSON KL. Relationship between quadriceps strength and rate of loading during gait in women. *J Orthop Res* 2000, 18:171–175
15. LIIKAVAINIO T, ISOLEHTO J, HELMINEN HJ, PERTTUNEN J, LEPOLA V, KIVIRANTA I ET AL. Loading and gait symmetry during level and stair walking in asymptomatic subjects with knee osteoarthritis: Importance of quadriceps femoris in reducing impact force during heel strike? *Knee* 2007, 14:231–238
16. JEFFERSON RJ, COLLINS JJ, WHITTLE MW, RADIN EL, O'CONNOR JJ. The role of the quadriceps in controlling impulsive forces around heel strike. *Proc Inst Mech Eng H* 1990, 204:21–28
17. HUNT MA, HINMAN RS, METCALF BR, LIM BW, WRIGLEY TV, BOWLES KA ET AL. Quadriceps strength is not related to gait impact loading in knee osteoarthritis. *Knee* 2010, 17:296–302
18. ZHANG Y, McALINDON TE, HANNAN MT, CHAISSON CE, KLEIN R, WILSON PW ET AL. Estrogen replacement therapy and worsening of radiographic knee osteoarthritis: The Framingham study. *Arthritis Rheum* 1998, 41:1867–1873
19. HANNA FS, WLUKA AE, BELL RJ, DAVIS SR, CICUTTINI FM. Osteoarthritis and the postmenopausal woman: Epidemiological, magnetic resonance imaging, and radiological findings. *Semin Arthritis Rheum* 2004, 34:631–636
20. KELLGREN JH, MOORE R. Generalized osteoarthritis and Heberden's nodes. *Br Med J* 1952, 1:181–187
21. JANSSEN I, MARK AE. Separate and combined influence of body mass index and waist circumference on arthritis and knee osteoarthritis. *Int J Obes (Lond)* 2006, 30:1223–1228
22. DE KLERK BM, SCHIPHOF D, GROENEVELD FPMJ, KOES BW, VAN OSCH GJVM, VAN MEURS JBJ ET AL. Limited evidence for a protective effect of unopposed oestrogen therapy for osteoarthritis of the hip: A systematic review. *Rheumatology (Oxford)* 2009, 48:104–112
23. WANG Y, SIMPSON JA, WLUKA AE, TEICHTAHL AJ, ENGLISH DR, GILES GG ET AL. Relationship between body adiposity measures and risk of primary knee and hip replacement for osteoarthritis: A prospective cohort study. *Arthritis Res Ther* 2009, 11:R31
24. RAYMAKERS JA, SAMSON MM, VERHAAR HJJ. The assessment of body sway and the choice of the stability parameter(s). *Gait Posture* 2005, 21:48–58
25. MOIR GL. Three different methods of calculating vertical jump height from force platform data in men and women. *Meas Phys Educ Exerc Sci* 2008, 12:207–218
26. LIDDLE D, ROME K, HOWE T. Vertical ground reaction forces in patients with unilateral plantar heel pain – a pilot study. *Gait Posture* 2000, 11:62–66
27. GRABINER MD, FEUERBACH JW, LUNDIN TM, DAVIS BL. Visual guidance to force plates does not influence ground reaction force variability. *J Biomech* 1995, 28:1115–1117
28. RISKOWSKI JL, MIKESKY AE, BAHAMONDE RE, ALVEY TV 3rd, BURR DB. Proprioception, gait kinematics, and rate of loading during walking: Are they related? *J Musculoskelet Neuronal Interact* 2005, 5:379–387
29. BIANCO R, AZEVEDO APS, FRAGA CHW, ACQUESTA FM, MOCHIZUKI L, AMADIO AC ET AL. The influence of running shoes cumulative usage on the ground reaction forces and plantar pressure responses. *Rev Bras Educ Fís Esporte* 2011, 25:583–591
30. PUDDLE DL, MAULDER PS. Ground reaction forces and loading rates associated with parkour and traditional drop landing techniques. *J Sports Sci* 2013, 12:122–129.eCollection 2013
31. BRANDT KD, RADIN EL, DIEPPE PA, VAN DE PUTTE L. Yet more evidence that osteoarthritis is not a cartilage disease. *Ann Rheum Dis* 2006, 65:1261–1264
32. JONES GM, WATT DG. Muscular control of landing from unexpected falls in man. *J Physiol* 1971, 219:729–737
33. MESSIER SP, ETTINGER WH Jr, DOYLE TE, MORGAN T, JAMES MK, O'TOOLE ML ET AL. Obesity: Effects on gait in an osteoarthritic population. *J Appl Biomech* 1996, 12:161–172
34. BROWNING RC, KRAM R. Effects of obesity on the biomechanics of walking at different speeds. *Med Sci Sports Exerc* 2007, 39:1632–1641
35. TONGEN A, WUNDERLICH RE. Biomechanics of running and walking. *Mathem Spor* 2010, 315–328
36. WEYAND PG, STERNLIGHT DB, BELLIZZI MJ, WRIGHT S. Faster top running speeds are achieved with greater ground forces not more rapid leg movements. *J Appl Physiol (1985)* 2000, 89:1991–1999
37. SOL C, MITCHELL K, TÖRÖK DJ, BANKS SA, GRAVES S, WELSH R. Impact forces at the knee joint: A comparative study on running styles. *Med Sci Sports Exerc* 2001, 33:S128
38. MESSIER SP, LOESER RF, HOOVER JL, SEMBLE EL, WISE CM. Osteoarthritis of the knee: Effects on gait, strength, and flexibility. *Arch Phys Med Rehabil* 1992, 73:29–36
39. RADIN EL, YANG KH, RIEGGER C, KISH VL, O'CONNOR JJ. Relationship between lower limb dynamics and knee joint pain. *J Orthop Res* 1991, 9:398–405
40. TAYLOR WR, HELLER MO, BERGMANN G, DUDA GN. Tibio-femoral loading during human gait and stair climbing. *J Orthop Res* 2004, 22:625–632
41. FISHER NM, WHITE SC, YACK HJ, SMOLINSKI RJ, PENDERGAST DR. Muscle function and gait in patients with knee osteoarthritis before and after muscle rehabilitation. *Disabil Rehabil* 1997, 19:47–55
42. HEINO BRECHTER J, POWERS CM. Patellofemoral stress during walking in persons with and without patellofemoral pain. *Med Sci Sports Exerc* 2002, 34:1582– 1593
43. DILLON PZ, UPDYKE WF, ALLEN WC. Gait analysis with reference to chondromalacia patellae. *J Orthop Sports Phys Ther* 1983, 5:127–131
44. NADEAU S, GRAVEL D, HÉBERT LJ, ARSENAULT AB, LEPAGE Y. Gait study of patients with patellofemoral pain syndrome. *Gait Posture* 1997, 5:21–27
45. GRIFFITH JF. *Diagnostic ultrasound: Musculoskeletal.* Elsevier, Philadelphia, 2015

Corresponding author:

E. Paraskevopoulos, 28 Agiou Spiridonos street, 122 43 Egaleo, Attica, Greece
e-mail: elparaskevop@uniwa.gr