Journal of Biomechanics 48 (2015) 392–395

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech
www.JBiomech.com

Short communication

Inertial control as novel technique for in vitro gait simulations

T. Natsakis [a,*], J. Burg [a,b], G. Dereymaeker [a], I. Jonkers [b], J. Vander Sloten [a]

[a] KU Leuven, Department of Mechanical Engineering, Celestijnenlaan 300c, Box 2419, 2001 Heverlee, Belgium
[b] KU Leuven, Faculty of Kinesiology and Rehabilitation Science, Tervuursevest 101, Box 1500, 3001 Heverlee, Belgium

ARTICLE INFO

Article history:
Accepted 26 November 2014

Keywords:
Gait simulations
In vitro
Control theory
Total ankle prosthesis
Triple arthrodesis

ABSTRACT

In vitro gait simulations are a preferential platform to study new intervention techniques or surgical procedures as they allow studying the isolated effect of surgical interventions. Commonly, simulations are performed by applying pre-defined setpoints for the kinetics and kinematics on all degrees of freedom (DOFs) of the cadaveric specimen. This however limits the applicability of the experiment to simulations for which pre-defined kinematics and kinetics can be measured in vivo. In this study we introduce inertial control as a new methodology for gait simulations that omits the need for pre-defined setpoints for the externally applied vertical ground reaction force (vGRF) and therefore allows the effect of interventions to be reflected upon it. Gait simulations of stance (1 s) were performed in 10 cadaveric specimens under three clinically relevant conditions: native ankle, total ankle prosthesis (TAP) and total ankle prosthesis plus triple arthrodesis (TAP+TA). In the native ankle, simulated vGRF was compared against the vGRF measured in vivo in 15 healthy volunteers and high correlations were found ($R^2 = 0.956$, slope of regression line S=1.004). In TAP and TAP+TA, vGRF changed, therefore confirming the sensitivity of the method to kinematic constrains imposed with surgery. Inertial control can replicate in vivo kinetic conditions and allows investigating the isolated effect of surgical interventions on kinematic as well as kinetics.

1. Introduction

When studying surgical interventions on the foot, in vitro gait simulations allow isolating the effect of surgery on the resulting kinematics and kinetics (Bayomy et al., 2010; Suckel et al., 2007; Valderrabano et al., 2003; Weber et al., 2012). This is highly relevant as it is known from literature (Thomas et al., 2006; Wu et al., 2000; Valderrabano et al., 2007) that factors such as time since operation, pain, and muscle training, can influence the measured kinematics and kinetics in patients. Therefore, the isolated effect of a surgical intervention cannot be studied in vivo.

During current in vitro gait simulations, input set points derived from a control group are applied in all degrees of freedom (DOFs) of the specimen. These set-points are in the form of either tibial kinematics (Aubin et al., 2012; Noble et al., 2010; Sharkey and Hamel, 1998) or a combination of tibial kinematics and vGRF (Hurschler et al., 2003; Nester et al., 2007). This, however, limits the applicability to simulations where it is of interest to impose pre-defined kinematics and kinetics measured in vivo, e.g. when studying bone kinematics during normal gait. When however, the sole effect of an intervention on the kinematics or kinetics is studied, this approach cannot be used. The simulation is over-constrained and it does not allow the effect of the specific intervention to be reflected in the kinematics and kinetics as they are imposed and not measured.

To overcome this limitation, we present a new technique that alleviates the need for a predefined set-point for the vertical tibial kinematics or vGRF during in vitro gait simulations, which leaves one flexible DOF for the effect of the interventions to be reflected upon. To demonstrate the applicability of the technique, gait simulations were performed in intact cadaveric foot specimens and the resulting vGRF was evaluated. To motivate its clinical relevance, vGRF was also evaluated during gait simulations with identical input parameters but after applying a total ankle prosthesis (TAP) and total ankle prosthesis plus triple arthrodesis (TAP+TA) locking the hindfoot motion of the specimen.

2. Methods

The inertial control approach is applied on a custom built CGS that manipulates the sagittal plane tibial kinematics of cadaveric foot specimens (Fig. 1). Even though the remaining 3 DOFs are constrained, previous validation studies (Peeters et al., 2013; Natsakis et al., 2012) demonstrated that the CGS is able to reconstruct kinematics similar to those measured in vivo. To account for the effect of plantarflexion and the resulting up and downwards translation of the knee axis, a supporting plate parallel to the ground translates vertically and modifies the foot position. This motion is imposed by a pneumatic actuator (Festo ADNGF-63-100-P-A, Sankt Ingbert, Germany) operating in a force feedback loop. The force applied by the

* Corresponding author.

http://dx.doi.org/10.1016/j.jbiomech.2014.11.044
0021-9290/

actuator is determined by the inertial controller. A force plate (9281 B Kistler Multicomponent Force Plate, Kistler Instruments GmbH, Germany), mounted on top of the supporting platform, contacts the foot and measures the vGRF at 2 kHz. Stance phase is simulated in one second and the horizontal translation and sagittal plane rotation of the tibia as well as the activation of 6 muscle groups (peroneal muscles, tibialis anterior, tibialis posterior, flexor digitorum, triceps surae and flexor halucis) are controlled based on prescribed set-points. The vertical kinematics of the tibia imposed by the translating ground plate is defined using the inertial controller approach. All devices are operated and synchronised by a custom built controller programmed in LabVIEW 2013 (National Instruments, TX, USA).

In vivo, the (vGRF) exerted on each foot during walking is expressed as in Eq. (1) (Richards, 2008) (Fig. 2a), with m_h the mass of the person, $a_h(x)$ the vertical acceleration of the body's centre of gravity, W(x) the portion of the weight of the person supported by that foot and x the percentage of stance phase.

$$vGRF_{invivo}(x) = W(x) + m_h * a_h(x) \quad (1)$$

$W(x)$ is a function of the percentage of stance phase (x), with weight increasing from 0 to 100% body weight (BW) during the first 16% of the stance phase (initial double support (IDS)) and decreasing back to 0% BW, during the final 16% (terminal double support (TDS)) (Richards, 2008). The duration of the IDS and TDS was chosen for normal gait conditions, it can however be adapted to correspond to the value observed in a specific patient group. Assuming continuity and smoothness during this transition, the weight carried by the foot during stance phase is described by Eq. (2) where $f(x)$ is an algebraic sigmoid function with input x being the stance phase percentage (Eq. (3)) (Fig. 3).

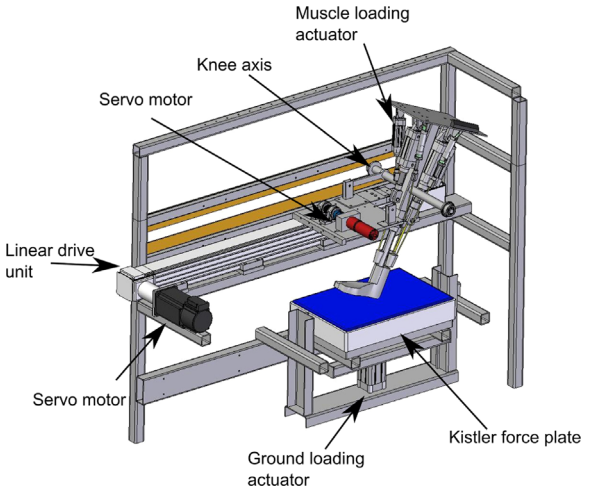

Fig. 1. Cadaveric Gait Simulator (CGS). In the figure, the motor imposing the horizontal motion (black), the sliding rail in the horizontal direction, the carriage with the actuators, the tibial rotation motor (red) and the supporting plate (blue) of the foot are presented. (For interpretation of the references to color in this figure caption, the reader is referred to the web version of this paper.)

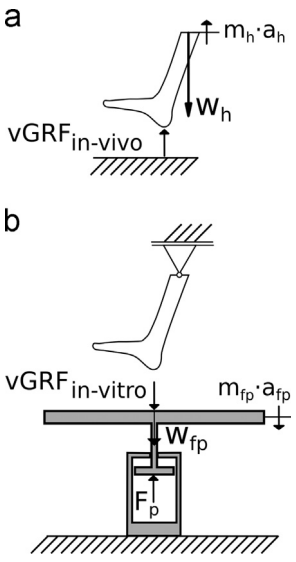

Fig. 2. Free body diagram of (a) the foot and (b) the supporting plate of the CGS. The force of the actuator (Fp), the weight of the plate (W_{fp}) and the reaction force of the foot ($vGRF_{invitro}$) impose acceleration (a_{fp}) on the mass of the plate (m_{fp}). The force delivered by the plate to the foot is equal in magnitude and opposite in direction to the force delivered by the foot to the plate ($vGRF_{invitro}$). The accelerations have different directions between in vitro and in vivo cases, as downward movement of the plate is equivalent to an upward movement of the knee.

$$vGRF_{invivo}(x) = W_h * f(x) + m_h * a_h(x) \quad (2)$$

$$f(n) = \begin{cases} \dfrac{0.5x^2}{x^2 - 184x + 128}, & \text{if } x < 16 \\ 1, & \text{if } 16 \le x \le 84 \\ \dfrac{0.5(x-100)^2}{x^2 - 184x + 8528}, & \text{if } x > 84 \end{cases} \quad (3)$$

Based on the free body diagram of the supporting plate of the CGS (Fig. 2b), Eq. (4) can be composed. The force (F_p) applied by the actuator on the plate is controlled to be equal to the weight of the plate (W_{fp}), plus a simulated weight of the specimen (W_{sw}) multiplied by the sigmoid function ($f(x)$) to simulate (vGRF) during double support Eq. (5). To ensure cadaveric integrity, the simulation weight was reduced to 50% of the control subject weight (49.7 kg), and a similar reduction was applied to the muscle forces ($W_{sw} = 240$ N). Combining Eq. (4) and (5) into (6), the (vGRF) profile corresponding to the in vivo situation (Eq. (2)) is simulated.

$$vGRF_{invitro}(x) - F_p(x) + W_{fp} = m_{fp} * a_{fp}(x) \quad (4)$$

$$F_p(x) = W_{fp} + W_{sw} * f(x) \quad (5)$$

$$vGRF_{invitro}(x) = W_{sw} * f(x) + m_{fp} * afp(x) \quad (6)$$

Following Eq. (6) the mass (m_{fp}) and acceleration (a_{fp}) of the force plate corresponds to the mass (m_h) and acceleration (a_h) of the body's centre of mass during normal walking. More specific, to mimic the upwards acceleration of the centre of mass during push off, the force plate will move down with the inertia of the supporting plate contributing to the (vGRF) during simulated gait.

To evaluate the proposed methodology and to demonstrate a clinically relevant example, in vitro gait simulations were performed in 10 freshly frozen cadaveric foot specimens using the inertial control approach under three different conditions. The feet were initially tested intact and afterwards, a three component total ankle prosthesis (TAP) (Hintegra, New Deal, Lyon, France) was implanted, followed by a triple arthrodesis between the bones of the hind foot (TAP+TA). All surgical procedures were performed by an experienced foot surgeon and 15 repetitions were performed for each condition. All repetitions were used during the analysis of the results.

The results were compared against vGRF that was captured in vivo in 15 healthy subjects, using two force plates (AMTI OR6 series, AMTI, MA, USA) at 1 kHz and normalised for BW. Since the weight of the person (W_h) is equal to the mass (m_h) multiplied by the gravitational acceleration ($g = 9.8$ m/s^2) (Eq. (7)), the normalised forces, based on Eq. (2), are of the form (Eq. (8)).

$$W_h = m_h * g \quad (7)$$

$$vGRF_{norm,invivo}(x) = \frac{W_h * f(x) + m_h * a_h(x)}{W_h} = f(x) + \frac{a_h(x)}{g} \quad (8)$$

For the in vitro measurements, a difference between the simulated weight ($W_{sw} = 240$ N) and the mass of the platform ($m_{fp} = 70$ kg) may arise. Therefore, a different normalisation approach was used. The simulated weight over stance phase function ($W_{sw} * f(x)$) was first subtracted from the measured vGRF and the result was divided by the mass of the force plate (m_{fp}) multiplied by the gravitational force (g). Finally, the sigmoid function $f(x)$ was added, so that it

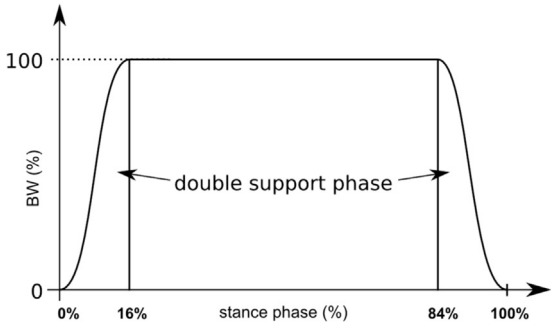

Fig. 3. The function f(x) of the weight carried by the foot during stance phase. The sigmoid parts are visible during the transition period of the double supported phases. The function is continuous and smooth over the whole duration of stance phase.

corresponded with the normalised in vivo measurements (Eq. (9)).

$$vGRF_{norm,invitro}(x) = \frac{vGRF_{dynamic,invitro} - W_{sw} * f(x)}{m_{fp} * g} + f(x) = f(x) + \frac{a_{fp}(x)}{g} \quad (9)$$

Using a non-parametric rank sum test (Wilcoxon) for each percentage of stance phase, the intact in vitro vGRF was compared against intact in vivo condition and the TAP in vitro and TAP+TA in vitro conditions were compared against intact in vitro. Statistically significant differences were reported for $p < 0.05$. A linear regression was finally performed within the mean values of intact in vivo and in vitro measurements. The statistical analysis was performed with R v3.0.2 (R Core Team, 2013).

3. Results

The normalised signals of the (vGRF) obtained from the in vitro simulations and in vivo gait analysis, are presented in Fig. 4. Timing of the significant differences is indicated using an asterisk (*). For the intact ankle, significant differences in vGRF of the intact in vivo and intact in vitro data are primarily found during the first 10% and between 80% and 90% of stance phase. Only isolated differences from 62% to 90% are found between the intact in vitro and TAP in vitro measurements. More significant differences are found comparing the intact in vitro and TAP+TA in vitro measurements around the first (13%–14%) and especially around the second peak (63%–92%). The regression between the intact in vitro and intact in vivo was $R^2 = 0.956$ with slope $S = 1.004$.

4. Discussion

Reproducing physiologic gait in cadaveric specimens poses many challenges and several research groups have developed custom built cadaveric gait simulators trying to address them (Whittaker et al., 2011; Noble et al., 2010; Nester et al., 2007; Sharkey and Hamel, 1998; Hurschler et al., 2003; Kim et al., 2001). In several designs, physiologic tibial kinematics must be applied, coupled with adequate force production to the different tendon actuators. These input variables must be synchronised and applied accurately. Furthermore, these signals must be appropriate for each cadaveric specimen, to account for differences in the geometry and condition (Natsakis et al., 2012). Finally, when the interest of a study is to investigate the isolated effect of a parameter (e.g. surgical intervention) the simulations should not be over-constrained to allow the phenomena that are studied to be reflected on the signals that are being measured.

In this study, we evaluate a methodology for performing gait simulations on cadaveric specimens without the need of applying predefined vGRF. We evaluate its sensitivity in two specific interventions and to isolate their effect, muscle forces remain unaltered across conditions. Gait was simulated in intact ankle, TAP and TAP+TA joint, allowing one DOF to adapt (vGRF and vertical kinematics) and controlling the remaining DOF (horizontal kinematics, tibial rotation) and muscle actuation with identical set-points across different conditions. We, therefore, assumed that the changes in the biomechanics of the foot introduced by the subsequent interventions would be reflected upon the measured vGRF, as this illustrates the sensitivity of this methodology.

The results from this methodology were compared against data captured during in vivo gait analysis. The high R^2 values ($R^2 = 0.956$) observed between the intact in vivo and intact in vitro group, indicates that the proposed methodology accurately reproduces physiological vGRF. The vGRF between the two groups was significantly different during limited part of stance phase i.e. just before the 1st and slightly after the 2nd peak. This however does not limit the validity of the methodology, as the magnitude of the differences is low and they appear during very short stance phase intervals. Furthermore, the changes in magnitude of the vGRF following the subsequent surgical procedures, demonstrates the sensitivity of the methodology to changes in the condition of the foot. The less pronounced differences between the intact in vitro and TAP in vitro than between intact in vitro and TAP+TA in vitro groups are introduced as the triple arthrodesis, that locks different midfoot joints, constrains the foot motion more with larger impact on the kinematics compared to TAP.

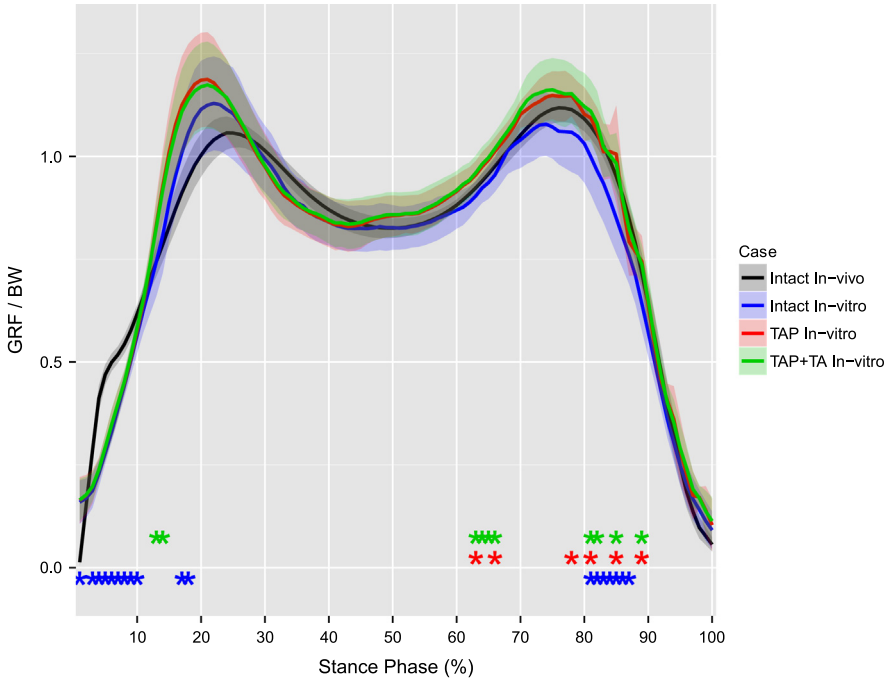

Fig. 4. Normalised vGRF of intact in vivo (black), intact in vitro (blue), TAP in vitro (red) and TAP+TA in vitro (green) groups. The average vGRF and the standard deviation of all feet and repetitions are presented. Significant differences between the intact in vivo and in vitro (blue), the intact in vitro and TAP in vitro (red) and the intact in vitro and TAP+TA in vitro (green) measurements are marked with an asterisk (*). (For interpretation of the references to color in this figure caption, the reader is referred to the web version of this paper.)

To the best of our knowledge, this study is the first presenting such a methodology, which opens new possibilities in in vitro gait simulations e.g. studying the effect of new implant design, muscle activation or surgical interventions when the effect of the intervention needs to be isolated and not affected by pain or fear of a patient. vGRF obtained from our CGS using the inertial controller shows equally high correspondence with vGRF obtained from in vivo gait analysis with those reported in other studies. Furthermore, the simulation speed is increased compared to previous designs, being very close to physiologic speed. Finally, with the need for a pre-defined set-point removed for the vGRF, the inertial controller provides opportunities to simulate gait under different conditions and investigate surgical interventions or new implant designs and their influence on gait parameters.

Conflict of interest statement

The authors have no conflicts of interest to report.

Acknowledgments

This work was funded by the Chair Berghmans–Dereymaeker, the Research Foundation Flanders and the Agency for Innovation by Science and Technology in Flanders (IWT). The authors would like to thank Pieter Spaepen (KU Leuven, Campus Groep T, Belgium) for his valuable feedback during the development of this methodology.

References

Aubin, P.M., Whittaker, E., Ledoux, W.R., 2012. A robotic cadaveric gait simulator with fuzzy logic vertical ground reaction force control. IEEE Trans. Robot. 28, 246–255. http://dx.doi.org/10.1109/TRO.2011.2164958, URL: ⟨http://ieeexplore.ieee.org/lpdocs/epic03/wrapper.htm?arnumber=6029994 ⟩.

Bayomy, A.F., Aubin, P.M., Sangeorzan, B.J., Ledoux, W.R., 2010. Arthrodesis of the first metatarsophalangeal joint: a robotic cadaver study of the dorsiflexion angle. J. Bone Jt. Surg. 92, 1754–1764. http://dx.doi.org/10.2106/JBJS.I.00984, URL: ⟨http://jbjs.org/article.aspx?articleid=6097⟩.

Hurschler, C., Emmerich, J., Wülker, N., 2003. In vitro simulation of stance phase gait part I: model verification. Foot Ankle Int. 24, 614–622. http://dx.doi.org/10.1177/107110070302400808, URL: ⟨http://www.ncbi.nlm.nih.gov/pubmed/12956567⟩.

Kim, K.J., Kitaoka, H.B., Luo, Z.P., Ozeki, S., Berglund, L.J., Kaufman, K.R., Kai-Nan, A., 2001. In vitro simulation of the stance phase in human gait. J. Musculoskelet. Res. 5, 113–121. http://dx.doi.org/10.1142/S0218957701000490.

Natsakis, T., Peeters, K., Burg, F., Dereymaeker, G., Vander Sloten, J., Jonkers, I., 2012. Specimen-specific tibial kinematics model for in vitro gait simulations. Proc. Inst. Mech. Eng. Part H J. Eng. Med. 227, 454–463. http://dx.doi.org/10.1177/0954411912464459, URL: ⟨http://pih.sagepub.com/lookup/doi/10.1177/0954411912464459⟩.

Nester, C.J., Jones, R.K., Liu, A., Howard, D., Lundberg, A., Arndt, A., Lundgren, P., Stacoff, A., Wolf, P., 2007. Foot kinematics during walking measured using bone and surface mounted markers. J. Biomech. 40, 3412–3423. http://dx.doi.org/10.1016/j.jbiomech.2007.05.019, URL: ⟨http://www.sciencedirect.com/science/article/B6T82-4P6MC1F-2/2/aa9061bd02e005aa68516f79ad887cff⟩.

Noble, L.D., Colbrunn, R.W., Lee, D.G., van den Bogert, A.J., Davis, B.L., 2010. Design and validation of a general purpose robotic testing system for musculoskeletal applications. J. Biomech. Eng. 132, 025001–0250012. http://dx.doi.org/10.1115/1.4000851, URL: ⟨http://www.ncbi.nlm.nih.gov/pubmed/20370251⟩.

Peeters, K., Natsakis, T., Burg, J., Spaepen, P., Jonkers, I., Dereymaeker, G., Van der Sloten, J., 2013. An in vitro approach to the evaluation of foot-ankle kinematics: performance evaluation of a custom-built gait simulator. Proc. Inst. Mech. Eng. H. 227, 955–967. http://dx.doi.org/10.1177/0954411913490455, URL: ⟨http://www.ncbi.nlm.nih.gov/pubmed/23736995⟩.

R Core Team, 2013. R : A Language and Environment for Statistical Computing. URL: ⟨http://cran.r-project.org/doc/manuals/fullrefman.pdf⟩.

Richards, J., 2008. Biomechanics in Clinic and Research: An interactive teaching and learning course, first ed. Churchill Livingstone/Elsevier, London, URL: ⟨http://www.biomechanicsonline.com/⟩.

Sharkey, N.A., Hamel, A.J., 1998. A dynamic cadaver model of the stance phase of gait: performance characteristics and kinetic validation. Clin. Biomech. 13, 420–433. http://dx.doi.org/10.1016/S0268-0033(98)00003-5, URL: ⟨http://www.sciencedirect.com/science/article/B6T59-3TP0GFR-6/2/d6566767e02e3fb28002a142f9167492⟩ ⟨http://www.sciencedirect.com/science?_ob=ArticleURL&_udi=B6T59-3TP0GFR-6&_user=10083421&_coverDate=09/30/1998&_rdoc=1&_fmt=high&_orig=search&_origin=search&_sort=d&_docanchor=&view=c&_acct=C000047079&_version=1&_urlVersion=0&_userid=10083421&md5=d9b85f191e61093268de55cf9a88fec2&searchtype=a⟩ ⟨http://www.scien?edirect.com/science?_ob=MImg&_imagekey=B6T59-3TP0GFR-6-4&_cdi=4997&_user=10083421&_pii=S026⟩.

Suckel, A., Muller, O., Herberts, T., Langenstein, P., Reize, P., Wulker, N., 2007. Talonavicular arthrodesis or triple arthrodesis: peak pressure in the adjacent joints measured in 8 cadaver specimens. Acta Orthop. 78, 592–597. http://dx.doi.org/10.1080/17453670710014275, URL: ⟨http://www.ncbi.nlm.nih.gov/pubmed/17966017⟩.

Thomas, R., Daniels, T.R., Parker, K., 2006. Gait analysis and functional outcomes following ankle arthrodesis for isolated ankle arthritis. J. Bone Jt. Surg. Am. 88, 526–535. http://dx.doi.org/10.2106/JBJS.E.00521, URL: ⟨http://www.ncbi.nlm.nih.gov/pubmed/16510818⟩.

Valderrabano, V., Hintermann, B., Nigg, B.M., Stefanyshyn, D., Stergiou, P., 2003. Kinematic changes after fusion and total replacement of the ankle: part 1: range of motion. Foot Ankle Int. 24, 881–887, URL: ⟨http://www.ncbi.nlm.nih.gov/pubmed/14733341⟩.

Valderrabano, V., Nigg, B.M., von Tscharner, V., Stefanyshyn, D.J., Goepfert, B., Hintermann, B., 2007. Gait analysis in ankle osteoarthritis and total ankle replacement. Clin. Biomech. (Bristol, Avon) 22, 894–904. http://dx.doi.org/10.1016/j.clinbiomech.2007.05.003, URL: ⟨http://www.ncbi.nlm.nih.gov/pubmed/17604886⟩.

Weber, J.R., Aubin, P.M., Ledoux, W.R., Sangeorzan, B.J., 2012. Second metatarsal length is positively correlated with increased pressure and medial deviation of the second toe in a robotic cadaveric simulation of gait. Foot Ankle Int. 33, 312–319. http://dx.doi.org/10.3113/FAI.2012.0312, URL: ⟨http://www.ncbi.nlm.nih.gov/pubmed/22735203⟩.

Whittaker, E.C., Aubin, P.M., Ledoux, W.R., 2011. Foot bone kinematics as measured in a cadaveric robotic gait simulator. Gait Posture 33, 645–650. http://dx.doi.org/10.1016/j.gaitpost.2011.02.011, URL: ⟨http://www.sciencedirect.com/science/article/pii/S0966636211000610⟩ ⟨http://www.sciencedirect.com/science?_ob=MiamiImageURL&_cid=271166&_user=10083421&_pii=S0966636211000610&_check=y&_origin=search&_coverDate=30-Apr-2011&view=c&wchp=dGLbVlt-zSkzS&md5=4111b6963bd320558c67ed7fd5761dde/1-s2.0-S0966636211000610-ma?n.pdf⟩.

Wu, W.L., Su, F.C., Cheng, Y.M., Huang, P.J., Chou, Y.L., Chou, C.K., 2000. Gait analysis after ankle arthrodesis. Gait Posture 11, 54–61, URL: ⟨http://www.ncbi.nlm.nih.gov/pubmed/10664486⟩.