Ambulatory Measurement of Ground Reaction Forces

Peter H. Veltink, *Member, IEEE*, Christian Liedtke, Ed Droog, and Herman van der Kooij

***Abstract*—The measurement of ground reaction forces is important in the biomechanical analysis of gait and other motor activities. Many applications require full ambulatory measurement of these forces, but this is not supported by current measurement systems. We propose the use of two six-degrees-of-freedom force and moment sensors under each shoe, which enables the ambulatory measurement of ground reaction forces and centers of pressure (CoP). The feasibility of this method is illustrated by experimental results in a healthy subject, using a force plate as a reference. The ground reaction forces and CoP recordings show good correspondence when they are evaluated for forces above 40 N and when it is simply assumed that the sensors are flat on the ground when they are loaded. The root mean square (rms) difference of the magnitude of the ground reaction force over 12 gait trials was 15 ± 2 N, corresponding to $1.9 \pm 0.3\%$ of the maximum ground reaction force magnitude. The rms difference of the horizontal component of the ground reaction force was 3 ± 2 N, corresponding to $0.4 \pm 0.2\%$ of the maximum ground reaction force magnitude and to $2 \pm 1\%$ of the maximum of the horizontal component of the ground reaction force. The rms distance between both CoP recordings is 2.9 ± 0.4 mm, corresponding to $1.1 \pm 0.2\%$ of the length of the shoe, when the trajectories are optimally aligned.**

***Index Terms*—Biomechanics, center of pressure (CoP), ground reaction force, instrumentation, legged locomotion, sensing.**

I. Introduction

THE ACCEPTED measurement system for biomechanical assessment of human ambulation today is a laboratory-bound system with optical measurement of movement and force plate measurement of ground reaction forces (GRF) [1]. The force plates, commonly one or two, are fixed in the floor of the gait laboratory. From the body movements and ground reaction forces measured with these laboratory systems, joint moments can be estimated by inverse dynamics methods [1], [2]. A biomechanical model of the body is required for this analysis.

In this laboratory setting, force plates seriously limit the evaluation of mobility performance. In most cases, one or, at the most, two plates are available in a gait laboratory, fixed in the floor. First, this requires subjects to place their feet completely on the force plate in order to perform a correct measurement of the total ground reaction force. This is, in fact, an extra restriction for walking which is undesired and especially difficult for patients with gait impairments. Second, only one or two steps during a gait trial can be measured, while there is, in general, a large variation in the ground reaction forces between steps, related with differences in muscle activation and body movement. In order to characterize the variability in the ground reaction force patterns and to determine their mean and standard deviation, many steps need to be analyzed. With force plates, this requires many trials of which only one or two steps are measured. Third, when standing with both feet on a single force plate, the ground reaction force of each foot cannot be distinguished; the plate only measures the total ground reaction force. Fourth, the use of fixed force plates impedes ambulatory ground reaction measurement in the daily-life environment during daily-life activities at home and at work. Ambulatory assessment is, however, required in many applications of human movement analysis, including the evaluation of the impact of rehabilitation treatments in daily life [3], [4] and the ergonomic evaluation of working tasks and environments [5]. Laboratory systems are not suitable for such evaluations. Alternative movement analysis systems that can be applied in an ambulatory setting have been developed, including inertial and magnetic sensory systems [6]–[9]. However, no suitable ambulatory GRF measurement systems are available.

The only ambulatory alternatives to force plates used to our knowledge are pressure sensor matrices, placed inside the shoe [10]. These matrices only allow the estimation of the vertical component of the GRF, not the shear components. In addition, pressure measured by pressure sensor matrices applied inside the shoe, does not add up to the GRF under the shoe, because of the pressure induced by the fitting of the shoe. A method to estimate GRF using insole pressure sensors and additional knowledge of body movements has been proposed [11], but an independent measurement of GRF is preferred. Miniature sensors that can measure all components of stress inside a shoe have been proposed [12], but are not yet used in regular human gait analysis. Recently, also carpets of pressure sensor matrices on the floor have been used [10]. This allows more than two steps to be measured but is, like force plates, not an ambulatory system.

It is the objective of the current study to design and evaluate a new method for ambulatory measurement of all components of the ground reaction force and center of pressure for each foot separately. The major requirements for such a system are small weight and size, and no impediment of functional mobility.

II. Methods

We propose to use two six-degrees-of-freedom force sensors: one under the heel and the other under the forefoot (Fig. 1). The calculation of the GRF and centers of pressure (CoP) from the sensor signals is first presented. Subsequently, the experimental methods for evaluating the performance of this instrumented force shoe are described.

Manuscript received July 16, 2004; accepted December 2, 2004. This work was supported in part by the Dutch Ministry of Economic Affairs (ICT breakthrough project ExoZorg) and in part by the Dutch Foundation for Scientific Research (NWO) (LOPES project).

The authors are with the Institute for Biomedical Technology (BMTI), University of Twente, 7500 AE Enschede, The Netherlands (e-mail: P.H.Veltink@utwente.nl).

Digital Object Identifier 10.1109/TNSRE.2005.847359

Fig. 1. Shoe instrumented with two 6-axis force and moment sensors.

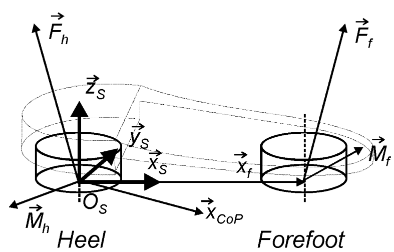


Fig. 2. Definition of sensor coordinate system S linked to the heel sensor of the instrumented shoe.

A. Calculation of GRF and Center of Pressure

All three components of the GRF $\vec{F}_{GR}$ were estimated by summing the forces measured by the heel and forefoot sensors

$$^{S}\vec{F}_{GR} = {}^{S}\vec{F}_h + {}^{S}\vec{F}_f. \tag{1}$$

Subscript h indicates the heel sensor, f the forefoot sensor. The forces were expressed in the coordinate system S, defined at the interface between the shoe and the ground and aligned with the orientation of the shoe (Fig. 2). This coordinate system was renewed for each foot placement to coincide with the heel sensor when the heel was on the ground. The origin O_S coincides with the point on the axis of the heel sensor, which is lying on the interface plane with the floor when the heel is on the ground. The unit coordinate vector $\vec{x}_S$ was chosen to point toward the intersection point of the axis of the forefoot sensor and the interface plane with the floor, the unit coordinate vector $\vec{z}_S$ was chosen vertical, and the coordinate vector $\vec{y}_S$ was chosen such that the resulting global coordinate system $\{\vec{x}_S, \vec{y}_S, \vec{z}_S\}$ was right-handed. The floor was assumed to be flat and horizontal.

The CoP was calculated as being the position ${}^{S}\vec{x}_{\mathrm{CoP}}$, which satisfies the following conditions.

1. The position is on the ground: ${}^{S}x_{\mathrm{CoP},z} = 0$.
2. The x and y components of the moment ${}^{S}M(\vec{x})$ exerted on the shoe equal zero: ${}^{S}M_x({}^{S}\vec{x}_{\mathrm{CoP}}) = 0$ and ${}^{S}M_y({}^{S}\vec{x}_{\mathrm{CoP}}) = 0$.

The moment ${}^{S}M(\vec{x})$ exerted at any position ${}^{S}\vec{x}$ can be written as follows (Fig. 2):

$${}^{S}\vec{M}(\vec{x}) = {}^{S}\vec{M}_h + {}^{S}\vec{M}_f + (-{}^{S}\vec{x}) \times {}^{S}\vec{F}_h + ({}^{S}\vec{x}_f - {}^{S}\vec{x}) \times {}^{S}\vec{F}_f. \tag{2}$$

TABLE I
MAIN SPECIFICATIONS OF THE SIX-DEGREES-OF-FREEDOM FORCE AND MOMENT SENSOR (ATI FTD-MINI45-SI-580–20) USED FOR AMBULATORY GRF MEASUREMENT

Sensor parameter	value
Diameter	45 mm
Thickness	15.7 mm
mass	92 g
Vertical force range	± 1160 N
Horizontal force range	± 580 N
Moment range (in all 3 directions)	± 20 Nm

Where ${}^{G}\vec{x}_f$ is the position vector of the center of the forefoot sensor expressed in coordinate system S, the center of the forefoot sensor being defined as the intersection point of the axis of the forefoot sensor and the interface plane with the floor.

This results in the following coordinates of the CoP:

$$\begin{aligned} {}^{S}x_{\mathrm{CoP,x}} &= -\frac{{}^{S}M_{h,y} + {}^{S}M_{f,y} - {}^{S}x_{f,x}{}^{S}F_{f,z}}{{}^{S}F_{h,z} + {}^{S}F_{f,z}} \\ {}^{S}x_{\mathrm{CoP,y}} &= \frac{{}^{S}M_{h,x} + {}^{S}M_{f,x} + {}^{S}x_{f,y}{}^{S}F_{f,z}}{{}^{S}F_{h,z} + {}^{S}F_{f,z}} \\ {}^{S}x_{\mathrm{CoP},z} &= 0. \end{aligned} \tag{3}$$

B. Experimental Methods

The right shoe of a pair of scandals (size 42) was instrumented with two six degrees of freedom force/moment sensors (Fig. 1). The center of the sensors were at a distance of 192 mm. Two aluminum dummies were applied under the left shoe. The specifications of the applied sensors (ATI-Mini45-SI-580–20, supplier: Schunk, Arnhem, NL) are given in Table I.

A healthy subject wearing the instrumented shoes walked repeatedly over an AMTI force plate. Each of the 12 trials consisted of three or four strides, of which one was on the force plate. All force and moment signals were sampled at 120 samples/s. Before further processing, all signals of the instrumented shoe sensors and force plate were low-pass filtered at a cutoff frequency of 30 Hz using a zero phase digital filter, applying a second-order Butterworth filter twice, in both the forward and reverse directions (MATLAB *filtfilt* function). GRF and center of pressure as a function of time (CoP trajectory) were calculated from the signals measured by the force plate, as well as by the instrumented shoes and compared. The signals of the force sensors were only analyzed when the sensors were loaded, assuming that they were flat on the ground in that phase. This condition occurred in the first half of the stance phase for the

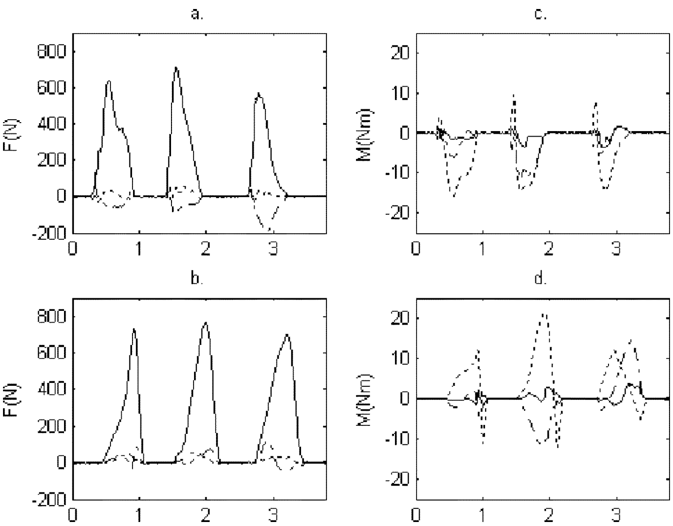


Fig. 3. Reaction forces and moments measured using the instrumented shoe in a representative trial, expressed in the coordinate system S. a: Heel sensor force. b: Forefoot sensor force. c: Heel sensor moment. d: Forefoot sensor moment. x: Dash–dotted. y: Dotted. z: Solid.

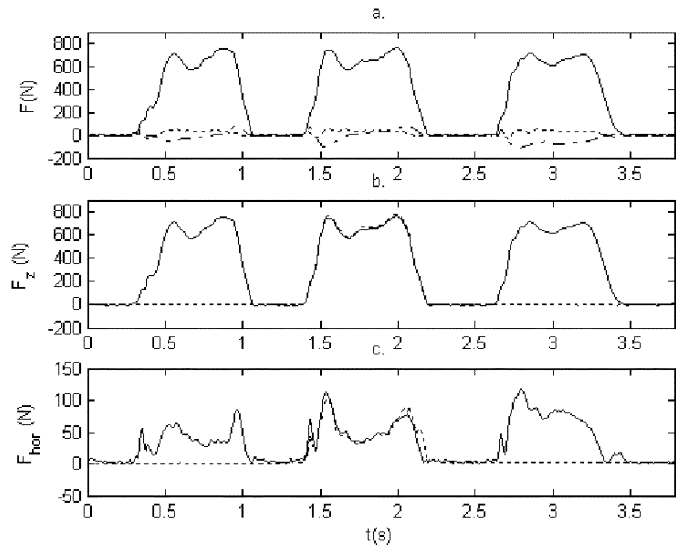


Fig. 4. GRF measured by instrumented shoe and force plate (same trial as in Fig. 3; the second step is on the force plate).a: GRF measured by the instrumented shoe, expressed in coordinate system S (x: dashed; y: dotted; z: solid) b: Vertical component of the GRF measured by the instrumented shoe (solid) and force plate (dotted). c: Horizontal component of the GRF measured by the instrumented shoe (solid) and force plate (dotted).

heel sensor and the second half of the stance phase for the forefoot sensor and was evaluated using a threshold of 40 N to the force measured by the sensors. The force plate signals were transformed to the coordinate system S by applying a rotation around the vertical axis and a displacement in the horizontal plane that optimally aligned both CoP trajectories, minimizing their rms distance. For this purpose, the *fminunc* routine from the MATLAB optimization toolbox was used, applying the rms distance between both CoP trajectories as the optimization criterion.

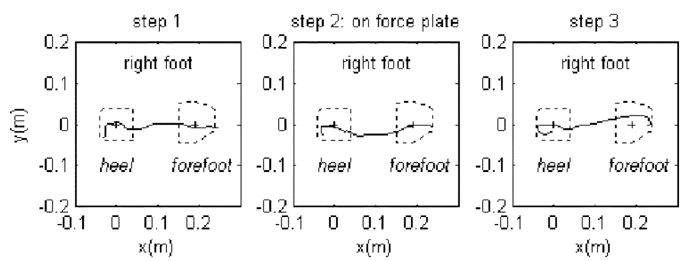


Fig. 5. CoP of three steps measured by the instrumented shoe and expressed in the coordinate system S (same trial as in Fig. 3). Second step is on the force plate. Dashed line indicates the borders of the support surfaces under the sensors. CoP was analyzed for ground reaction forces above a threshold of 40 N.

III. RESULTS

The forces and moments measured by the heel and forefoot sensors during the three steps of the right foot in a representative trial are presented in Fig. 3. The total ground reaction force was calculated by adding the forces of both sensors [(1); Fig. 4(a)]. The close resemblance of the ground reaction force measured by the instrumented shoe and the force plate is illustrated in Fig. 4(b) and (c): Fig. 4(b) presenting a comparison of the vertical and Fig. 4(c) of the horizontal component of the GRF. The rms difference of the GRF magnitude estimates was 15 ± 2 N over the 12 evaluated trials, being $1.9 \pm 0.3\%$ of the maximal GRF magnitude. The rms difference of the estimates of the horizontal component of the GRF was 3 ± 2 N ($0.4 \pm 0.2\%$ of the maximal GRF magnitude or $2 \pm 1\%$ of the maximal horizontal component of the GRF). Fig. 4 illustrates the variability of the ground reaction force measured between steps, indicating the importance of analyzing all steps instead of only one or two, as is commonly done when using force plates. The variability of the CoP of subsequent steps is illustrated in Fig. 5. The CoP estimates stay within the borders of the support surfaces under the sensors. For ground reaction forces below 40 N, the start and end of the CoP trajectories would be estimated outside of the support surfaces, indicating that the assumption that the sensors were flat on the ground did not apply below this threshold. It should be noted, however, that the time the force is between zero and 40 N is only approximately 5% of the stance phase time and the associated impulse is below 0.2% of the total impulse of the stance phase [see e.g., Fig. 4(b)]. The comparison of the optimally aligned CoP estimates from the instrumented shoe and the force plate is illustrated in Fig. 6. The CoP trajectories agree well, resulting in an rms distance between both CoP trajectories of 2.9 ± 0.4 mm over all 12 trials, corresponding to $1.1 \pm 0.2\%$ of the length of the shoe. The x and y components of the force plate GRF in the coordinate system S were estimated using the same rotation that aligned the CoP trajectories. They compared favorably with the x and y components calculated from the instrumented shoe signals, as illustrated in Fig. 7. Over all 12 trials, the rms difference of the x component was 19 ± 3 N, corresponding to $2.3 \pm 0.3\%$ of the maximal GRF magnitude or $18 \pm 2\%$ of the maximal x component. The rms difference of the y component was 11 ± 3 N, corresponding to $1.4 \pm 0.4\%$ of the maximal GRF magnitude or $14 \pm 3\%$ of the maximal y component. It should be noted that these rms differences are considerably larger than the rms difference of the horizontal component of the GRF presented above, indicating a remaining error in the rotation applied to the GRF of the force plate. Corresponding to GRF and CoP (Figs. 4 and 5), the x and

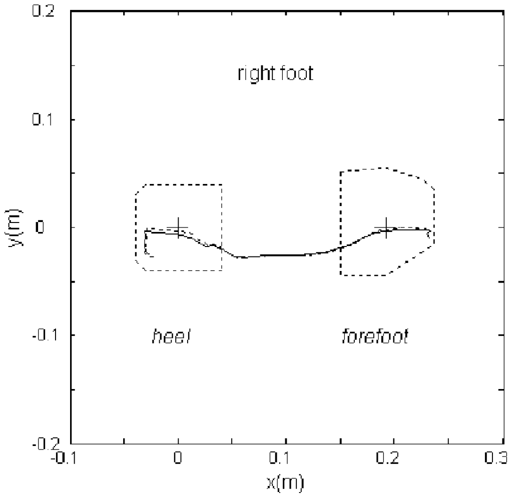


Fig. 6. CoP of the step on the force plate measured by the instrumented shoe (solid) and by the force plate (dashed), both expressed in the coordinate system S (same trial as in Fig. 3). To this end, the force plate CoP trajectory was optimally aligned with the CoP trajectory of the instrumented shoe, using a displacement and rotation in the horizontal plane, minimizing rms distance of the CoP positions (resulting minimal rms distance in this trial was 2.3 mm). CoP was analyzed for ground reaction forces above a threshold of 40 N.

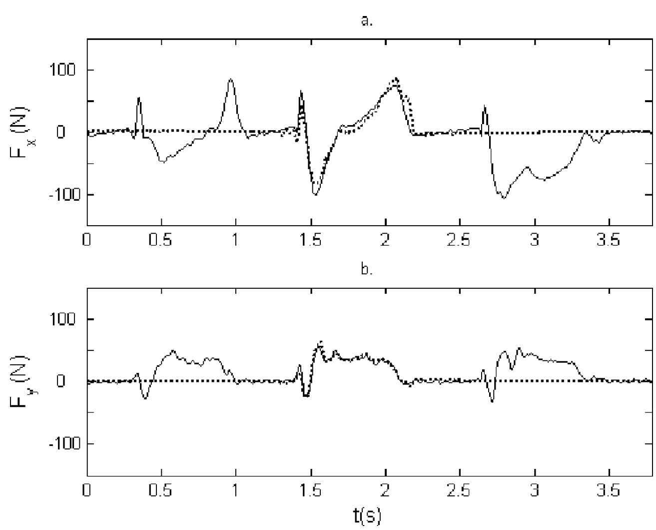


Fig. 7. x and y components of the GRF (a and b, respectively) measured by the instrumented shoe (solid) and the force plate (dotted) for the same trial as in Fig. 3. Note that the second step was on the force plate. Horizontal components of the GRF measured by the force plate were transformed to the coordinate system S by rotation over the angle that resulted in optimal alignment of the CoP trajectories of the force plate and instrumented shoe (see Fig. 6).

y components of the GRF also show considerable variation from step to step, emphasizing the need for measurement of all steps. The x component of the third step shows the deceleration at the end of the four-step trial, while the step on the force plate shows the standard deceleration and acceleration components in x direction known from gait at constant speed.

IV. Discussion

The results indicate that ambulatory measurement of the ground reaction force under each foot is feasible when using two six-degrees-of-freedom force sensors per shoe.

The proposed measurement system allows complete measurement of the ground reaction forces during daily-life activities at home and at work. In combination with ambulatory movement analysis methods reported in recent years [6]–[9], [13], this may allow for full biomechanical analysis of movement tasks at any place and outside of a movement analysis laboratory. It should be noted that this biomechanical analysis may still be possible if the subject exerts force via other parts of the body; for example, when manipulating the environment using the hands. The additional unmeasured reaction forces may be estimated when taking into account the biomechanical characteristics of the body (segment masses and dimensions) and the conditions for assuring body balance.

The rms distance between the CoP trajectories obtained with both measurement systems after optimal alignment was small. The remaining difference may be due to the assumption that the sensors were flat on the ground when measuring substantial force. The rms error may be even lower when this orientation change is taken into account. The larger rms errors in the x and y components of the GRF in comparison to the horizontal component indicates a remaining error in the rotation angle used for aligning the CoP trajectories. The actual transformation should be investigated by measuring the movement of the shoe, for example using an optokinetic measurement system and reflective markers on the heel and forefoot sensors.

The advantage of the instrumented shoes lies in the fact that the sensors are moving with the foot, potentially giving a lower error in estimating the relative position between foot and sensor. In addition, this error is not influenced by the dimensions of the applied measurement volume, as is the case with optokinetic measurement systems. It should be noted, however, that additional errors will occur also with the instrumented shoes, when estimating forces and moments in coordinate systems of body segments and joints, because of uncertainties in relative positions and deformation of segments, especially of the foot. This needs to be further evaluated.

The proposed sensor arrangement allows normal or near-normal gait because the sole of the shoe can be flexed during push-off. The main indications of near normal gait are the normal biphasic shape of the ground reaction force and the normal timing of the force patterns. The current study, however, does not provide a thorough evaluation of the normality of the gait pattern. This can be assessed by comparing walking on the instrumented shoes with walking over a force plate in several types of normal shoes.

In principle, ground reaction forces can be measured using a single sensor per foot. However, this would require a rather rigid plate under the sensor for generating the necessary moments associated with possible CoP trajectories, while avoiding that the sole of the shoe touches the ground directly at any place. Such a rigid plate would impede normal gait.

The proposed design of the instrumented shoe with two force sensors can still be improved, for example, by optimizing the compliance of the sole above the sensors and of the plates under the sensors. Both sole and plates under the sensor do not need to be very stiff for valid measurements of ground reaction forces. It is essential that the complete ground reaction force is measured by the sensors and, thus, that the sole does not touch the ground directly. However, the application of more flexible ground plates would require constant measurement of the orientation of the sensors, which is in principle possible using inertial sensors, but would make the measurement system more complex.

A second possible optimization of the design would be the reduction of the total height of the ground plate, the sensor, and the sole. The sensor, having a height of 15.7 mm, could be partly integrated in the sole and the thickness of the ground plate could be minimized. In the current design, this has not been optimized. For ease of experimentation and safety of the sensors, additional connection plates above and under the sensors were used in the current study, increasing the effective height.

The measurements in this study illustrate the need for assessment of all steps during gait. In order to balance the body during gait, each step needs to be different [14]. The analysis of the variability of gait, and thus the measurement of all steps, is even more important in patients with neuromuscular disorders like stroke [15].

REFERENCES

[1] D. A. Winter, *Biomechanics and Motor Control of Human Movement*. New York: Wiley, 1990.

[2] B. Koopman, H. J. Grootenboer, and H. J. De Jong, "An inverse dynamics model for the analysis, reconstruction and prediction of bipedal walking," *J. Biomech.*, vol. 28, pp. 1369–1376, 1995.

[3] J. B. J. Bussmann, P. H. Veltink, F. Koelma, R. C. v. Lummel, and H. J. Stam, "Ambulatory monitoring of mobility-related activities: The initial phase of the development of an activity monitor," *Eur. J. Phys. Med. Rehabil.*, vol. 5, pp. 2–7, 1995.

[4] H. B. Buschmann, P. J. Reuvekamp, P. H. Veltink, W. L. Martens, and H. J. Stam, "Validity and reliabiltiy of measurements obtained with an "activity monitor" in people with and without a transtibial amputatation," *Phys. Therapy*, vol. 78, pp. 989–998, 1998.

[5] C. T. M. Baten, P. Oosterhoff, I. Kingma, P. H. Veltink, and H. J. Hermens, "Inertial Sensing in ambulatory load estimation," in Proc. 18th Annu. Int. Conf. IEEE Eng. Med. Biol. Soc., Amsterdam, The Netherlands, Oct. 31–Nov. 3 1996.

[6] A. T. Willemsen, J. A. van Alste, and H. B. Boom, "Real-time gait assessment utilizing a new way of accelerometry," *J. Biomech.*, vol. 23, pp. 859–63, 1990.

[7] E. R. Bachman, "Inertial and magnetic tracking of limb segment orientation for inserting humans into synthetic environmnets," Ph.D. dissertation, Naval Postgrad. School, Monterey, CA, 2000.

[8] H. J. Luinge, "Inertial sensing of human movement," Ph.D. dissertation, Univ. Twente, Enschede, The Netherlands, 2002.

[9] H. J. Luinge and P. H. Veltink, "Inclination measurement of human movement using a 3-D accelerometer with autocalibration," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 12, no. 1, pp. 112–121, Mar. 2004.

[10] J. Woodburn, "The contribution of plantar pressure measurment to the understanding of foot structure and function in rheumatoid arthritis," in Proc. Biomechanics Lower Limb Health, Disease Rehabil., Salford, U.K., Sep. 1–3, 2003, pp. 90–92.

[11] A. Forner Cordero, "Human gait, stumble and . . . fall?—Mechanical limitations of the recovery from a stumble," Ph.D. dissertation, Univ. Twente, Enschede, The Netherlands, 2003.

[12] M. A. Razian and M. G. Pepper, "Design, development, and characteristics of an in-shoe triaxial pressure measurment transducer utilizing a single element of piezoelectric copolymer film," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 11, no. 3, pp. 288–293, Sep. 2003.

[13] P. H. Veltink, P. Slycke, J. Hemssems, R. Buschman, G. Bultstra, and H. J. Hermens, "Three dimensional inertial sensing of foot movements for automatic tuning of a two-channel implantable drop-foot stimulator," *Med. Eng. Phys.*, vol. 25, pp. 21–28, 2002.

[14] C. E. Bauby and A. D. Kuo, "Active control of lateral balance in human walking," *J. Biomech.*, vol. 33, pp. 1433–1440, 2000.

[15] D. Roetenberg, J. H. Buurke, P. H. Veltink, A. Forner-Cordero, and H. J. Hermens, "Surface electromyography analysis for variable gait," *Gait Posture*, vol. 18, pp. 109–117, 2003.

Peter H. Veltink (S'85–M'88) was born in Groenlo, The Netherlands, in 1960. He received the M.Sc. degree (cum laude) in electrical engineering and the Ph.D. degree for his research in the area of electrical nerve stimulation from the University of Twente, Enschede, The Netherlands, in 1984 and 1988, respectively.

Currently, he is a Professor of technology for the restoration of human function at the Institute for Biomedical Technology (BMTI), the University of Twente. He performs research in the area of artificial motor control and ambulatory sensory systems with applications to rehabilitation medicine. He has been the Scientific Coordinator of three European Union (EU) research training networks and is and has been involved in various projects financed by the EU, the Dutch ministry of Economic Affairs, and the Dutch Foundation for Technical Sciences (STW). He performed sabbaticals at Case Western Reserve University, Cleveland, OH, in 1989 and at the Center for Sensory-Motor-Interaction, Aalborg University, Aalborg, Denmark, in 1997.

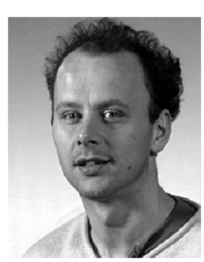
Prof. Veltink was the Treasurer of the International Functional Electrical Stimulation Society (IFESS) from 1996 to 2001. He received the Royal Shell Stimulating Prize for his contribution to the rehabilitation-engineering field in 1997.

Christian Liedtke received the Dipl. Ing. degree from the University for Applied Sciences, Ulm, Germany, in 2002. He is currently working toward the Ph.D. degree at the University of Twente, Enschede, The Netherlands.

His current research activities concern ambulatory monitoring of balance performance.

Ed Droog was born in Delft, The Netherlands, in 1954.

After receiving training in electronic engineering, he worked as a Technician in a hospital for more than four years. During the last 20 years, he has been a Technician with the Department of Biomedical Signals and Systems (BSS), Faculty of Electrical Engineering, Mathematics and Computer Science, the University of Twente, Enschede, The Netherlands. His area of expertise is electronic instrumentation.



Herman van der Kooij was born in Rotterdam, The Netherlands, in 1970. He received the M.Sc. degree in mechanical engineering and the Ph.D. degree (cum laude) in the area of human balance control from the University of Twente, Enschede, The Netherlands, in 1995 and 2000, respectively.

Currently, he is an Assistant Professor of biomechatronics at the Institute for Biomedical Technology (BMTI), the University of Twente, and performs research in the area of human motor control, rehabilitation robots, and virtual reality with applications to rehabilitation medicine.