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Review paper

Techniques for measuring weight bearing during standing and walking

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Abstract

Objective. To classify and assess techniques for measuring the amount of weight bearing during standing and walking.

Background. A large variety of weight bearing measuring techniques exists. This review describes their advantages and limitations to assist clinicians and researchers in selecting a technique for their specific application in measuring weight bearing.

Methods. A literature search was performed in Pubmed-Medline, CINAHL, and EMBASE. Measurement techniques were classified in 'clinical examination', 'scales', 'biofeedback systems', 'ambulatory devices' and 'platforms', and assessed on aspects of methodological quality, application, and feasibility.

Results. A total of 68 related articles was evaluated. The clinical examination technique is a crude method to estimate the amount of weight bearing. Scales are useful for static measurements to evaluate symmetry in weight bearing. Biofeedback systems give more reliable, accurate and objective data on weight bearing compared to clinical examination and scales, but the high costs could limit their use in physical therapy departments. The ambulatory devices can measure weight bearing with good accuracy and reliability in the hospital and at home. Platforms have the best methodological quality, but are mostly restricted to a gait laboratory, need trained personnel, and are expensive.

Conclusions. The choice of a technique largely depends upon the criteria discussed in this review; however the clinical utilisation, the research question posed, and the available budget also play a role. The new developments seen in the field of 'ambulatory devices' are aimed at extending measuring time, and improved practicality in data collection and data analysis. For these latter devices, however, mainly preliminary studies have been published about devices that are not (yet) commercially available.

Keywords: Measurement; Technique; Weight bearing; Standing; Walking; Rehabilitation

1. Introduction

Weight bearing during standing leads to a force exerted by gravity on the subject. In a standing position the amount of this vertical ground reaction force under both feet equals the weight of the subject. During walking, the vertical ground reaction force has a characteristic sinusoid shape during stance phase with two peak forces (Wittle, 1991). The amplitude of these peak forces correlates with the walking speed (Munro et al., 1987) and stride length (Martin and Marsh, 1992), and during 'normal' walking and running ranges from 0 to 5 times the body weight (Nigg, 1999; Nilsson and Thorstensson, 1989). It is evident that these forces under the foot during standing and walking generate forces and moments in other structures of the lower extremity, such as the hip (Bergmann et al., 1989; Davy et al., 1988).

In rehabilitation the amount of weight bearing during standing and walking is crucial in the healing period of orthopaedic patients with various pathologic conditions of the lower extremity, such as uncemented total hip

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arthroplasty, osteotomies, fractures of the leg, or amputees (Aranzulla et al., 1998; Chow and Cheng, 2000; Cunningham et al., 1989; Endicott et al., 1974; Gapsis et al., 1982; Karlsson et al., 1996; King et al., 1972; Lachiewicz et al., 1989; Phillips et al., 1991; Siebert, 1994) because immobilisation, non weight bearing, or excessive weight bearing can lead to complications (Bloomfield, 1997; Buehler et al., 1999; Convertino et al., 1997; Eisele et al., 2001; Teshima et al., 1992; Wirtz et al., 1998). Measurement of weight bearing is also essential during rehabilitation of patients with neurological pathologies, e.g. stroke, Parkinson, hemiplegia. and patients with diabetes mellitus and peripheral neuropathy to evaluate symmetry in weight bearing, weight shifting ability, and adequate limb loading (Bohannon et al., 1989; Bohannon and Tinti-Wald, 1991; Bohannon and Kelly, 1991; Kljajic and Krajnik, 1987; Mueller et al., 1994; Stokes et al., 1975; Wannstedt and Craik, 1978; Wannstedt and Herman, 1978). Assessment of the amount of weight bearing is, therefore, important.

Different techniques are used to measure the amount of weight bearing, corresponding to their field of application. Two major fields of application can be distinguished. First, the field of training patients to learn and control partial weight bearing (clinical measurement) (Gapsis et al., 1982; Chow et al., 1992; Engel et al., 1983; Gray et al., 1998; Perren and Matter, 1996; Tveit and Karrholm, 2001; Warren and Lehmann, 1975) and, second, the field of evaluating postoperative weight bearing (research measurement) (Hermens et al., 1986; Hynd et al., 2000; Kershaw et al., 1993; Koval et al., 1998). Within these two fields of application a large variety in measurement techniques and instruments exists, each with their advantages and limitations.

The purpose of this study is. therefore, to classify and assess the different techniques for measuring the amount of weight bearing on the lower extremity during standing and walking. This overview may assist clinicians and researchers to select the most suitable technique for their specific application in measuring weight bearing. The results may also indicate new directions in the development of techniques or instruments for measuring weight bearing.

2. Method

A literature search in Pubmed-Medline (1970–2001), CINAHL (1982–2001) and EMBASE (1990–2001) was performed using the following keywords: 'weight', 'bearing', 'load', 'force', 'foot', 'measure', 'walking aid' and, 'ambulant', 'platform' and 'device'. Of the articles generated, the reference lists were used to find other related articles. Only articles in the English and German language were selected.

2.1. Classification and definition of the measurement techniques

To compare the clinical measurement techniques a distinction was made in *clinical examination*, *scales*, and *biofeedback systems*. For the same purpose the research measurement techniques were classified in *ambulatory devices* and *platforms*. The following definitions were given to each of the clinical and research measurement techniques.

Clinical examination: Clinical examination was defined as observation and/or physical examination of the subject by a therapist during standing and walking with a walking aid, without extra instrumentation. Estimation of weight bearing during walking is done by observation, and/or palpation of the musculus triceps brachii, and/or by placing a hand under the foot of the affected leg.

Scales: Standard (analogue or digital) bathroom scales.

Biofeedback systems: Load monitoring systems that provide immediate feedback to the subject at a prescribed load level.

Ambulatory devices: Portable instruments with sensors attached to the subject, which allow continuous measurement. A division can be made between 'semi-portable' devices which use a long cable, and 'real' portable devices which allow unrestricted movement in the environment. Sensors can be placed under the bare foot, in the shoe, under the shoe, or in an insole.

Platforms: Instruments placed in or on the floor, or in a treadmill, for measuring the ground reaction force in one or more planes.

2.2. Assessment of measurement techniques

Each clinical and research measurement technique was assessed in order to compare aspects of methodological quality, application, and feasibility for measuring weight bearing.

For the methodological quality, information regarding the validity and reliability of the techniques was searched for in the literature. The quantitative criteria used in the literature related to validity were (Gray et al., 1998):

error: the difference between measured output value, measured by the system, and the true output value provided by a gold standard, mostly a force platform
accuracy: the true output value minus the measured output, divided by the true output value (this ratio is usually expressed in percent)
precision: the number of distinguishable alternatives from which a given result is selected, e.g. 2.400 N is more precise than 2.4 N (a high precision does, however, not imply a high accuracy)
drift: an undesirable change in output value, over time, during a constant input. When the

environmental factors temperature and humidity cause an undesirable change in output it is called *temperature* or *humidity* drift, respectively
hysteresis: the maximum deviation between ascending and descending output readings taken at the same input value
non-linearity: any deviation of the input–output characteristic from a straight line
creep: the ability of insole material used to resist change under an applied load/pressure over time, quantified as the difference between output value (force) and input value (force) divided by the input value (force) (McPoil et al., 1995; Woodburn and Helliwell, 1996).

Reliability or *reproducibility* was defined as the extent to which the instrument yields the same output on repeated measurements with equal input; reproducibility does not imply accuracy (Sim and Arnell, 1993).

Aspects related to application were: performance of 'static' (during standing) and/or 'dynamic' (during walking) measurements; 'maximum measurement time': duration of one measurement (e.g. one or multiple steps measured in one session); 'maximum time resolution': maximum sample frequency; and measurement range: the range in which the measurement variable (Newton, % body weight) can be measured.

The main aspects regarding feasibility were: simplicity of the technique to measure weight bearing for both the researcher (e.g. time needed, data transfer, data storage) and the subject (e.g. weight/size of the system), and the costs related to the measurements and/or purchase of the technique (information from sales literature).

3. Results

3.1. Clinical measurement techniques

3.1.1. Clinical examination

The *clinical examination* technique was used in a study by Gray et al. (1998) in which the physical therapist estimated what 60 pounds of force felt like when applied to the therapist's hand. They concluded that the amount of weight placed on the therapist's hand is subjective guesswork at best. No studies were found in which the validity of the clinical examination technique to measure weight bearing during standing or walking was determined. Also, no information was found on aspects of application and feasibility of the clinical examination technique.

3.1.2. Scales

The *scale* technique, which provides a quantitative outcome (kilogram) for the amount of weight bearing during standing, is less subjective than the clinical examination technique. Measurement on a standard (bathroom) scale ranges from 0 to approximately 130–150 kg, which usually allows to determine the amount of weight during double leg or single leg standing. Information on the accuracy of measuring the amount of weight for *scales* was reported by Winstein et al. (1996). They mentioned an accuracy of ±0.45 kg being the smallest unit of the analogue scale; the measurement range of the scales was 0–157.5 kg. The digital scales used by Bohannon et al. (1989) and Bohannon and Tinti-Wald (1991), were reported to register weight to a 0.1 pound (0.05 kg) level of precision, and were calibrated before each testing session. Chow and Cheng (2000) stated that accuracy was difficult to achieve and maintain particularly when the weight to be replicated was minimal. As scales are used to measure weight bearing in a static situation, Chow et al. (1992) found that the most consistent method to measure the actual weight under the feet during walking was a row of eight bathroom scales on the floor between parallel bars, but no data were given to confirm this statement.

3.1.3. Biofeedback systems

The first reported biofeedback device was a leg load warning system developed by Endicott et al. (1974) in 1974. It consisted of a single load transducer located in the hollowed-out heel of an orthopaedic sandal, and an electronic package that modified the signal from the sensor. It transmitted two audible tones, a low frequency tone when a patient had not exceeded the minimum load and a high-frequency tone when the maximum load was exceeded. The authors described good characteristics for the transducer, e.g. no hysteresis and no drifting of the signal over a period of 8 h, although no specific data were given on the methodological quality.

Miyazaki and Iwakura (1978a) made a limb load alarm device also using two different audio feedback signals. The low frequency tone was activated when the load exceeded a preset lower threshold and was switched off when the load increased to the upper threshold activating the high-frequency tone. Two strain gauge force transducers were attached to the sole of a shoe, one at the metatarsal part and at the heel, by elastic bands and Velcro straps. The device had a 10% accuracy, a temperature drift of ±3 N/°C. and a measurement range of 0–1000 N. The processing unit with rechargeable batteries ($12 \times 8 \times 2.5$ cm, 240 g) was fastened at the waist by a belt, allowing the system to work for 15 h. The costs were about $25, for components only. Limitation of the system was that to obtain the total limb load, by summing the outputs of both transducers, it was necessary that the remaining parts of the sole did not make contact with the floor; this was not feasible in all parts of the stance phase. However, the authors stated that the error due to this was small. In 1986 Miyazaki et al. (1986)

described a new limb load monitor mentioning the drawbacks of their previous system (Miyazaki and Iwakura, 1978a) in terms of accuracy and ease of setting the threshold load levels. This time they used a capacitive transducer, which resembled an insole and was also attached outside the shoe. The accuracy was improved to a 5% error. The device, however, was heavier (320 g) and larger ($13\times9\times3$ cm). A major problem of the system was that gain adjustment had to be made each time the transducer was changed, and to solve this problem expensive pre-calibrated load sensors were needed (Miyazaki et al., 1986).

The audio feedback system most referred to is the (Krusen) Limb Load Monitor (LLM) (Gapsis et al., 1982; Wannstedt and Craik, 1978; Kathrins and O'Sullivan, 1984; Wolf and Binder-MacLeod, 1982). The LLM consists of a pressure transducer built in an insole connected to a control box, which can be worn around the waist. The box emits a tone that may increase or decrease in pitch depending on loading calibration and mode selection. A control knob for adjusting the sound "null" point indicates to the patient that the desired loading has been reached. To calibrate the LLM, the patient loads the limb on a bathroom scale while the null point setting is adjusted (Wannstedt and Craik, 1978). Wolf and Binder-MacLeod (1982) compared the LLM with a force plate and found statistical significant differences ranging from 8% to 36% for both force peaks. The loading measurements showed a wide range in the 95% confidence intervals and it was therefore concluded that the accuracy of the LLM was insufficient. Intrarater and interrater reliability were determined by Carey et al. (1984). They found intra-class correlation coefficients of 0.995 and 0.990 for the first and second force peak, respectively, and concluded that the measurement reliability was high. However, healthy subjects were used in the study and intrarater and interrater reliability are highly dependent on individual peak forces and other gait variables. The authors therefore concluded that additional studies with different patient groups were necessary to establish the clinical utility of the LLM. Wolf and Binder-MacLeod (1982) described the LLM to be inexpensive and easy to use. However. Gapsis et al. (1982) had some criticism regarding the clinical usefulness, i.e. durability, and ease of repair, and opined that certain modifications of the LLM unit would increase the usefulness.

A PMT feedback system with insoles, based on a hydraulic principle, was used by Perren and Matter (1996). The device had a storage capacity of 8000 steps, and software presented the total amount of weight bearing in percentages of 100 N force units. Perren and Matter found that separate recording of the three sensors led to measurement errors. They also described having many technical problems, especially with the durability of these insoles. Siebert (1994) used a similar feedback device. Size and weight of the system were found to be acceptable. It was carried on the body and stored the weight load of each step which guaranteed the complete registration of the postoperative period of total hip patients. No data were given on methodological quality of the device.

Instrumented walking aids were designed by Bergmann et al. (1979) and Engel et al. (1983) to train and control weight bearing. Bergmann et al. chose for an indirect measuring technique which was inexpensive and not restricted to a certain place compared to the direct measurement techniques: the platforms and devices with transducers in or outside the shoe. They presented a linear relationship, between the ground reaction force of the walking aid and the partial weight bearing leg, to calculate the amount of weight bearing from the measurement of the walking aid. When the restriction in weight bearing is less than 10% of the body weight this technique becomes less reliable. According to Bergmann et al. the instrumented walking aid was easy to use and inexpensive. Engel et al. described a cane with a vibrating membrane built into the handle, and with two lights which can only be seen by the therapist. Although Engel et al. reported that their instrumented walking cane accurately indicated the amount of weight borne on the affected leg, these data were not validated against data from e.g. a force plate.

3.2. Research measurement techniques

3.2.1. Ambulatory devices: Semi-portable with transducers outside the shoe

A commercially available system (CDG) with eight capacitive transducers, covering the surface of the sole in almost every situation, was developed by Hermens et al. (1986). This semi-portable system (cable) measured the vertical force (distribution) during a walk of 20 s and was designed to be used in the clinical environment. The force transducers, placed in an overshoe, could be easily attached and removed form the patient's shoe. Hermens et al. mainly described the data processing procedures and methods of data presentation and not the methodological quality of the obtained force data (Table 1). The system was used in two clinical trials to measure the vertical ground reaction force during the gait of hip arthroplasty patients walking with crutches (Hesse et al., 1999; Sonntag et al., 2000).

3.2.2. Ambulatory devices: Semi-portable with transducers built in the shoe

The measuring system of Kljajic and Krajnik (1987) included five pairs of leather shoes in which eight or nine force transducers were built into each shoe sole. The accuracy (3%) was found to be comparable with force plates of the same cost. The advantage of this system over the force plate was that it enabled measurements of

Table 1
Aspects on the methodological quality of the ambulatory devices

System	Transducer type	Error	Creep	Drift	Hysteresis	Non-linearity	Reliability	Calibration
Commercial								
Pedar (McPoil et al., 1995; Boyd et al., 1997; Kalpen and Seitz, 1994; Kernozek et al., 1996; Quesada et al., 1997; Xia et al., 1994; Barnett et al., 2000)	Capacitive	0.8–17%	3.4%	<0.05 N/(cm^2 °C)[a,b]	<3%	?	0.84–0.98[c] 0.6–3%[d]	Air bladder device
F-Scan (McPoil et al., 1995; Woodburn and Helliwell, 1996; Quesada et al., 1997; Xia et al., 1994; Chen and Bates, 2000; Koch, 1993; Rose et al., 1992; Ahroni et al., 1998)	FSR	4–24%	11.6–19%	?	21%	?	0.52–0.98[c] 9.5–20.8%[d]	Subject's weight; Air bladder device
Parotec (Chesnin et al., 2000)	Piezoresistive	?	?	−0.001 N/cm^2[e] −0.015 N/cm^2/K[a]	0.05%	0.42%	?	By manufacturer
CDG (Hermens et al., 1986)	Capacitive	?	?	?	?	?	?	?
Non-commercial								
Tveit and Karrholm (2001)	Strain gauge	?	?	?	?	?	?	Subject on force plate
Aranzulla et al. (1998)	Resistive	<1kg	?	?	?	?	?	Testing machine
Abu-Faraj et al. (1997)	Conductive polymer	7–14%	?	?	5–10%	?	?	Device: dynamically at 36 °C
Dingwell et al. (1997)	Capacitive	?	?	?	?	?	?	Subject on force plate
Whalen et al. (1993)	Capacitive	?	?	?	?	?	?	Subject on force plate
Wertsch et al. (1992)	Conductive polymer	?	?	−0.5%/°C[a]	8%	Yes, ?	?	Device: at 36 °C
Zhu et al. (1991)	Conductive polymer	?	?	−0.5%/°C[a]	8%	Yes, ?	?	Device: dynamically at 36 °C
Gross and Bunch (1988)	Piezoelectric	3.1–9.9%	?	?	3.7%	2.3%	?	Device: dynamically
Kljajic and Krajnik (1987)	Strain gauge	3%	?	?	1%	1%	?	Subject on force plate
Miyazaki et al. (1986)	Capacitive	5%	?	?	?	Yes, ?	?	?
Hennig et al. (1982)	Piezoelectric	?	?	?	<1%	<2%	?	?
Miyazaki and Iwakura (1978b)	Strain gauge	10–15%	?	?	?	?	?	?

? = no data found in the literature.
[a] Temperature drift.
[b] Information from standard sales literature (2001).
[c] ICC.
[d] CV.
[e] Humidity drift.

a large number of steps, which the authors mentioned as being of utmost importance in severely impaired patients who cannot undergo the long and exhausting measurements required by force plate testing. Disadvantages were that the patient had to wear special shoes instead of his own footwear, and that measurements were restricted to a walkway.

3.2.3. Ambulatory devices: Semi-portable with transducers in insole

When measuring vertical forces with insoles a distinction can be made between discrete sensor insoles and matrix insoles. With discrete sensor insoles a limited amount of sensors are placed at specific areas under the foot, whereas matrix insoles consist of numerous sensors elements arranged in rows and columns which, unlike discrete systems, can measure the pressure/force under the entire plantar surface.

Gross and Bunch (1988) compared the vertical force output of their discrete insole system with a force plate to assess the transducer placement validity. They concluded that the shapes of the force curve were similar, but the units of load differed. The differences between curve endpoints were related to limited number of transducers available for placement beneath the calcaneus and toes. As no description was given of the duration of data storage or use of a data storage card we assume this system is semi-portable.

One of the first matrix insoles was developed by Hennig et al. (1982) using 499 piezoelectric transducers in each insole. Besides good sensor characteristics the sampling rate can be set up to 200 Hz. A limitation of the system is that the subject needs to carry a relatively large and heavy backpack, and external power via a cable was necessary because adequate batteries would be too heavy.

3.2.4. Ambulatory devices: Portable with transducers outside the shoe

Miyazaki and Iwakura (1978b) developed in 1978 a portable device, which could measure the vertical force under the foot continuously during standing and walking. It consisted of two strain gauge force transducers attached to the rear and front part of the sole of a sport shoe. Problems of the device were that a portion of the foot forces bypassed the force transducers due to direct contact between the floor and the sole of the shoe. At slow cadences (under 110 steps/min) the error was within 10%, but at higher cadences the error was 15%. The walking style of the subject was little affected due to the arrangement and thickness of the transducers, and the transducers varied in their sensitivity. Positive aspects were the long measurement time (8 h) and the relatively small and light weighted unit, which was fastened at the waist by a belt. Using a radio frequency for data transfer, with a transmission range from 15 to 100 m, no cable was needed and therefore the patient's movement was not restricted. To solve the aforesaid problems, Miyazaki and Ishida (1984) developed a new device consisting of two large flexible capacitive transducers per shoe. Specifically for patients with a fractured leg, Aranzulla et al. (1998) developed an ambulatory system which could continuously measure the amount of weight bearing for over 24 h. They used four flexible resistance transducers, which were attached to a Tubigrip sock. Custom-made software was used to calculate the mean amount of weight bearing, the mean duration of weight bearing, and the number of weight bearing events.

3.2.5. Ambulatory devices: Transducers in insole

A portable microprocessor-based data-acquisition system developed by Zhu et al. (1991) consisted of seven pressure sensors (0.5 mm) which were each dynamically calibrated at 36 °C after being placed in the insole to compensate for non-linearity and temperature drift. Data could be continuously collected for 7 min at a 20 Hz sample rate. The measurement time was extended by Wertsch et al. (1992) and Abu-Faraj et al. (1997) to 2 h with a sample frequency of 20 Hz, and to 8 h at 40 Hz, respectively. Abu-Faraj et al. described that the discrete sensors had thin metal backings which offered a greater stiffness than the rest of the insole by at least a 20:1 ratio. Another aspect was the insole distortion around the sensor edges. A positive aspect of discrete systems (because of their limited use of sensors compared to matrix insoles) is the ability of long-term data collection. Although the system was smaller but heavier than the one used by Zhu et al. and Wertsch et al., Abu-Faraj et al. described it as fully portable, not disrupting the natural gait pattern, and therefore ideal to measure the vertical force during daily living activities. To acquire data of a subject, customised insoles need to be made for each foot.

Whalen et al. (1993) developed a force measuring system with one capacitance insole force sensor designed to operate continuously for two weeks without the need to retrieve data or replace batteries. Long-term (two weeks) sensor stability, i.e. no significant change in the sensor force output over time, remained to be determined because after a 15-h trial the sensor failed due to two short bouts of activity (running and tennis). Data reduction was achieved by filtering the digitised vertical ground reaction force. The processor continuously time-differentiated the force and saved the maximum load rate between each peak and valley. The data logger stored the time of occurrence of these peaks and valleys and the total daily duration of force levels into 0.1 body weight intervals. Dingwell et al. (1997) replicated the device of Whalen et al. and measured four subjects for 10 h to quantify daily load bearing activity. Specific Matlab (The Mathworks, Inc.) routines were developed to convert the raw data to percent body weight (%BW). Tveit and Karrholm (2001) specially made pressure-sensitive insoles to measure the amount of weight bearing after hip surgery. Although no time interval was given, the system enabled long-term collection of data from each patient in his or her environment. The authors stated that further research would be required to evaluate patient compliance and long-term reliability. A commercially available discrete insole system is the Parotec system (Chesnin et al., 2000; Schaff, 1993). Chesnin et al. (2000) presented bench testing data for the methodological quality of the system with an accuracy of 2.0% and precision of 0.4%, and no discernible drift. The system is portable and can store data for 5 min with a sampling frequency up to 200 Hz.

Commercially available matrix insoles systems are the Pedar system (McPoil et al., 1995; Boyd et al., 1997;

Kalpen and Seitz, 1994; Kernozek et al., 1996; Quesada et al., 1997; Xia et al., 1994; Barnett et al., 2000) (Novel*gmbh*, Munich, Germany) (Barnett et al., 2000) and the F-Scan system (McPoil et al., 1995; Woodburn and Helliwell, 1996; Quesada et al., 1997; Xia et al., 1994; Chen and Bates, 2000; Koch, 1993; Rose et al., 1992) (Tekscan Inc., Boston, MA, USA) (Rose et al., 1992; Ahroni et al., 1998). In contrast to the previous described ambulatory devices, much information was found regarding the validity and reliability of measuring pressure/force by these two devices. A comparison between the two devices was made by McPoil et al. (1995) where the Pedar system demonstrated a high level of validity and reliability, whereas the results raised serious questions regarding the ability of the F-Scan insole to accurately measure normal force. Woodburn and Helliwell (1996) stated that accuracy of the F-Scan was hampered by inaccurate calibration, and poor hysteresis and poor creep properties. Quesada et al. (1997) compared the two systems after two new developments of the F-Scan, i.e. new resistive ink sensor insoles and software allowing calibration via an air pressure bladder (like the Pedar system) instead of the subject's body mass. They concurred with previous reports that the Pedar system is likely the system of choice when the greatest accuracy and repeatability are desired. Both systems use a cable for data transfer but the Pedar system also has a Mobile version which can collect data for up to 1 h on a 40 Mb PCMCIA storage card when both insoles, with 99 sensors each, and a sample rate of 50 Hz are used. The durability of the F-Scan sensor was criticised by Rose et al. (1992) and Woodburn and Helliwell (1996). The sensors showed consistent measurements for about 30 gait cycles but then the recordings steadily dropped off due to wear of the individual sensor. However, compared to the Pedar system, F-Scan has a higher measurement range, a higher sample frequency and the price of the system is lower (Table 2).

3.2.6. Standard platforms

The force plate is one of the most important measuring devices in biomechanics, quantifying external forces during human locomotion (Nigg, 1999). As with ambulatory devices, the electromechanical properties of the transducer used in the measuring instrument are of major importance for the quality of the output. The type of sensors used in force platforms, e.g. piezoelectric (Brennwald et al., 1974; Cobb and Claremont, 1995) or strain gauge (Beierlein, 1977; Ctercteko et al., 1981; Hutton and Drabble, 1972), have very good characteristics resulting in a high accuracy, and precision of force measurements made by these instruments (Nigg, 1999; Cobb and Claremont, 1995) (Table 3). The Kistler (Kistler Instrumente AG Winterthur, Switzerland) and AMTI (Advanced Mechanical Technology, Inc., Watertown, MA, USA) force platforms are, due to their characteristics, frequently used as a gold standard against which other systems are evaluated (Kalpen and Seitz, 1994; Barnett et al., 2000; Chen and Bates, 2000; Cobb and Claremont, 1995; Hargreaves and Scales, 1975). Hughes et al. (1991) stated that the reliability (Sim and Arnell, 1993) of the results depends on the capacity of the equipment to give the same result on consecutive steps, and on the ability of the patient to walk in the same way several times. Therefore, 100% reproducibility cannot be expected when measuring a variable related to gait, because gait always varies slightly between walks and subjects. When measuring weight bearing with standard floor platforms, multiple walks are needed to gain reliable results (Hughes et al., 1991; Bates et al., 1983; Edwards, 1986). Bates et al. (1983) used a Kistler platform and found that a minimum of eight trials was needed to obtain reasonable reliable mean vertical force values. However, multiple trials, needed to gain reliable results, can present problems for patients who have a poor physical condition (Kljajic and Krajnik, 1987; Mizrahi et al., 1985). Moreover, the subject needs to hit the force plate correctly, and a measurement protocol is needed to collect data in a standardised way (Cavanagh and Ulbecht, 1994; Quaney et al., 1995). The walking pattern will probably be affected due to targeting of the subject's foot on the platform, especially when the platform has small dimensions (Bergmann et al., 1979). Grabiner et al. (1995), however, found that variability of ground reaction force is not significantly affected by targeting the force plate. Wearing et al. (2000) also confirmed this by demonstrating that targeting a certain 30×24 cm^2 target does not affect ground reaction forces when a gait protocol that fine-tunes the start position is employed. The location of platforms is restricted to a gait laboratory because the platform generally needs to be mounted into the floor (Brennwald et al., 1974; Ctercteko et al., 1981; Hutton and Drabble, 1972; Simkin, 1981; Stauffer et al., 1974).

Portable platforms do not need to be mounted into the floor but need to be placed in a sufficiently long (6 m) walkway (Hughes et al., 1991; Quaney et al., 1995). Hughes et al. (1991) studied the reliability of the EMED F system (Novel*gmbh*) and found that a minimum of three trials was needed to obtain excellent reliability (Table 3). This system uses capacitance transducers and therefore has a relatively low (70 Hz) sample frequency. The AccuGait, a portable platform developed by AMTI, has a sample frequency up to 200 Hz. Generally, the time resolution of force platforms is high compared to ambulatory devices with sample frequencies >100 Hz (Tables 2 and 4).

3.2.7. Long platforms

To avoid the earlier mentioned problems of targeting and fatigue of patients, long force platforms were

Table 2
Aspects on application and feasibility of the ambulatory devices

System	Type	Measurement range (unit)	Time resolution	Weight/size	Data transfer	Data storage	Costs
Commercial							
Pedar (McPoil et al., 1995; Kernozek et al., 1996; Barnett et al., 2000)	Insole, matrix	0–60 (N/cm^2)	50 Hz	850 g/ 17.5×10.4×4.4 cm[a]	Data logger/ 10 m cable	1 h/40 Mb PCMCIA card/ PC	$14,230[b]
F-Scan (McPoil et al., 1995; Woodburn and Helliwell, 1996; Chen and Bates, 2000; Koch, 1993; Rose et al., 1992; Ahroni et al., 1998)	Insole, matrix	0–100 (N/cm^2)	165 Hz	180 g	9.25 m cable	PC	$12,950[b]
Parotec (Chesnin et al., 2000; Schaff, 1993)	Insole, discrete	0–62 (N/cm^2)	100–200 Hz	?	Data logger/ cable	5 min	$?
CDG (Hermens et al., 1986)	Overshoe, discrete		100 Hz	19×14×4.5 cm[a]	Cable	PC	$18,553[b]
Non-commercial							
Tveit and Karrholm (2001)	Insole, discrete	0–250 (kg)	250 Hz	?	Cable	PC	?
Aranzulla et al. (1998)	Tubigrip sock, discrete	? (kg)	?	500 g/ 13×13×7.5 cm	Data logger	24 h	?
Abu-Faraj et al. (1997)	Insole, discrete	0–1.2 (Mpa)	40 Hz	1250 g/ 15×15×10 cm	Data logger	8 h	?
Dingwell et al. (1997)	Insole, discrete	? (%BW)	25 Hz	908 g	Data logger	10 h/4 Mb PCMCIA card	?
Whalen et al. (1993)	Insole, discrete	? (%BW)	100 Hz	7.5×7.5×2.5 cm	Data logger	2 wks/2 Mb PCMCIA card	?
Wertsch et al. (1992)	Insole, discrete	0–2 (MPa)	100 Hz	800 g/ 20×18×17 cm	Data logger	2 h (20 Hz)	?
Zhu et al. (1991)	Insole, discrete	0–2 (Mpa)	100 Hz	800 g/ 20×18×17 cm	Data logger	7 min (20 Hz)	?
Gross and Bunch (1988)	Insole, discrete	0–2 (Mpa)	333 Hz	6×11×3 cm	?	?	?
Kljajic and Krajnik (1987)	In sole of shoe	(N)	100 Hz	?	30 m cable	PC	?
Miyazaki et al. (1986)	Sole outside shoe	0–1000 (N)	?	220 g/ 13×9×25 cm	Telemetry	PC	?
Hennig et al. (1982)	Insole, matrix	1–1.5 (Mpa)	200, 100, 50, 25 Hz	2900 g/ 25×18×15 cm	Cable	PC	?
Miyazaki and Iwakura (1978b)	Outside shoe	0–980 (N)	80 Hz	180 g/ 10.5×8×2.3 cm	Telemetry	PC	$70[c]

? = no data found in the literature.
[a] Information from standard sales literature (2001).
[b] Contains standard hardware and software (2001/2002).
[c] Costs at that time.

developed for clinical studies (Hynd et al., 2000; Olsson et al., 1986). Olsson et al. (1986) provided a walkway which consisted of two five-meter long platforms, developed in 1966 by Rydell, with additional equipment to give more accurate, efficient, fast and reliable data. The authors pointed out that this force plate walkway cannot record changes in forces that occur over 20 Hz due to its limited frequency range, nor can it study the highest frequencies of gait during initial foot contact. A specially designed computer program called ‘‘KI-step’’ calculated the maximal vertical force in %BW. Although the force plate walkway was designed for clinical studies it was not applicable to all types of gait disturbances. Olsson et al. stated that the subject must also be able to walk unassisted by another person, walk with one foot on each plate, and avoid placing the walking aid on the force plates. Hynd et al. (2000) developed a long dual-platform triaxial walkway. The natural frequency of the platform was 92 Hz, and thus above 50 Hz which is, according to the authors, sufficient for measurement of walking. The potential increase in walking base, as a result of having to walk on a different platform for each

Table 3
Aspects on the methodological quality of the platforms

System	Transducer type	Error	Creep	Drift	Hysteresis	Non-linearity	Reliability	Calibration
Standard platforms								
Kistler (type 9281)[a]	Piezoelectric	?	n.a.	?	<0.5%	<0.5%	?	By manufacturer static in situ
AMTI (type OR6-7)[a]	?	?	n.a.	0.01%/°C[b]	<0.2%	<0.2%	?	By manufacturer
EMED F (Hughes et al., 1991; Cavanagh and Ulbecht, 1994)	Capacitance	?	n.a.	0.5 kPa/°C[b]	<5%	?	0.75–0.90[c]	By manufacturer
AccuGait (AMTI)[a]	?	?	n.a.	?	?	?	?	By manufacturer
Long platforms								
Olsson et al. (1986)	Strain gauge	1%	n.a.	?	<1%	<1%	?	Static, discrete known loads
Hynd et al. (2000)	Strain gauge	0.1%	n.a.	?	0.1%	±0.1	?	Static, discrete known loads
Treadmill platforms								
Gaitway (Kistler)[a]	Piezoelectric	?	n.a.	?	See platform	See platform	?	By manufacturer static in situ
Kram and Powell (1989)	Strain gauge	1%	n.a.	?	?	<5%	?	Static, discrete known loads
Kram et al. (1998)	Strain gauge	1%	n.a.	?	?	0.2%	?	?
Belli et al. (2001)	Crystal force	?	n.a.	0.140 N/min	?	±0.3	?	Artificial leg method

? = no data found in the literature; n.a. = not applicable.
[a] Information from standard sales literature (2001/2002).
[b] Temperature drift.
[c] ICC.

Table 4
Aspects on application and feasibility of the platforms

System	Type	Measurement range	Time resolution	Weight/size	Data transfer	Data storage	Costs
Standard platforms							
Kistler (type 9281)[a]	In floor	20 kN[b]	1000 Hz	40 kg/60×40×10 cm	Cables	PC	$11,640[c]
AMTI (type OR6-7)[a]	In floor	17.8 kN[b]	570 Hz	32 kg/50.8×46.4×8.3 cm	Cables	PC	$11,640[c]
EMED F (Cavanagh and Ulbecht, 1994)	Portable		70 Hz[a]	43.8×22.6 cm[a]	Cables	PC	$9,299[a]
AccuGait (AMTI)[a]	Portable	2700 N	50, 100, 200 Hz	11.4 kg 50×50×4.4 cm	4.5 m cable	PC	$14,340[d]
Long platforms							
Olsson et al. (1986)	In floor	?	?	30 kg/500×20 cm	Cables	PC	?
Hynd et al. (2000)	In floor	2500 N	2000 Hz	35 kg/330×40 cm	Cables	PC	?
Treadmill platforms							
Gaitway (Kistler)[a]	Treadmill	2000 N, 6000 N	25–2500 Hz	364 kg/139×49.5×21.6[e]	Cables	PC	$44,545[c]
Kram and Powell, 1989	Treadmill	?	400 Hz	121×46 cm	Cables	PC	?
Kram et al. (1998)	Treadmill	?	1000 Hz	180×60 cm	Cables	PC	?
Belli et al. (2001)	Treadmill	?	800 Hz	200 m×25 cm	Cables	PC	?

? = no data found in the literature.
[a] Information from standard sales literature (2001/2002).
[b] Range vertical force.
[c] Platform only.
[d] Portable platform with amplifier and software.
[e] Bed length×width×height.

foot, was not apparent with the pathological gaits for which this platform walkway was designed.

3.2.8. Treadmill platforms

To measure the ground reaction force for many successive steps and with repeatable constant speed, Kram and Powell (1989) developed a treadmill-mounted force platform (Table 3). A commercially available strain gauged AMTI platform was mounted directly under the belt of a motorised treadmill. The 1.21 m long platform is usable up to a running speed of ~7 m/s. The limitation of this treadmill platform was that it could only measure the vertical ground reaction force. Kram et al. (1998), therefore, developed in 1998 a force treadmill that could also measure the horizontal ground reaction forces. They stated that the force treadmill has many advantages over conventional run-way mounted force platforms: it allows ground reaction forces to be collected far more rapidly, a large number of successive steps can be averaged to determine more representative values, no long laboratory or hallway is needed, and it allows for simultaneous collection of biomechanical and other data such as oxygen consumption and EMG. The treadmill ergometer developed by Belli et al. (2001) consists of two parallel treadmills, one for each foot. They used crystal force transducers (Kistler, Wintertur, Switzerland) because of their ability to tolerate wide range of force measurements. The maximum velocity of the treadmill belts is 2.87 m/s, which gives a theoretical maximum foot contact distance of 1.36 m. Comfortable walking requires a belt longer than the contact distance and therefore a 2 m long belt was used. Belli et al. found that low frequency oscillations were present in the force and velocity data collected during treadmill walking, and that future design should specify and reduce the velocity oscillations of the belt. The same phenomenon was also presented by Kram and Powell (1989); they concluded that, compared with a standard Kistler platform, the treadmill force platform provided accurate measurement of vertical ground force.

4. Discussion and conclusions

An extensive literature search was conducted on the different techniques and instruments used to measure weight bearing during standing and/or walking. Because of the large amount of different instruments we chose to classify the techniques into five categories in order to evaluate their advantages and disadvantages with regard to measurement of weight bearing. Not all instruments developed are mentioned in each category. For the *biofeedback systems* these are: the Accutread system (Chattanooga group, Hixson, USA), the PedAlert (Planet Products Corp., Madison, USA), the Biofeedback Weight Monitor (Enabling Devices, NY, USA), and the Andante Smartstep (Andante Medical Devices Ltd., Beer Sheva, Israel). For the *ambulatory devices* these are: the Footscan system (RS scan/International, Olen, Belgium), the Foot pressure system (T&T Medilogic, Schönefeld, Germany), and the Dinatto in-shoe pressure system (Buratto Advanced Technology, Crocetta, Italy). The main reason is that of these (commercially) instruments no articles were found in the used databases. We also found a large discrepancy in available articles between the five measurement techniques. Especially the limited information on clinical examination made it impossible to give a well-balanced conclusion on the methodological quality of this technique to measure weight bearing. In only a few of the found articles a weight bearing technique was used for a specific patient group. Therefore, we could not link a certain measurement technique to a particular clinical situation, e.g. fracture of the lower extremity or stroke. In most articles, in which a certain technique was evaluated, the authors described in the introduction paragraph that measurement of weight bearing is important in rehabilitation in general.

4.1. Clinical measurement techniques

Although *clinical examination* is a commonly used technique to train control weight bearing, it is a crude method estimating the amount of weight bearing during standing and walking, and allows the clinician to give only a qualitative (e.g. too high or too low) description of the outcome; such estimation results in a poor accuracy. Especially in a dynamic walking situation, where the magnitude depends on walking speed and stride length, assessment of the amount of vertical ground reaction force becomes even more complex. Only one article by Gray et al. (1998) was found in which the clinical examination technique was used. Although no comparison was made between the weight estimation by the clinician and e.g. a force platform, Gray et al. concluded that measuring the amount of weight by using the therapist's hand is "subjective guesswork at best".

Scales are often used in combination with clinical examination for training of weight bearing (Endicott et al., 1974; Siebert, 1994; Chow et al., 1992; Winstein et al., 1996). The measurement range of scales (0–150 kg) limits their use to static measurements. Scales have a good accuracy (Bohannon et al., 1989), but Chow and Cheng (2000) stated that accuracy is difficult to achieve and maintain, particularly when the weight to be replicated is minimal. The needle of a scale does not have peak and hold capacity so that it is difficult to read and reproduce peak forces (Gray et al., 1998). Therefore, this technique is mostly used for static measurements to evaluate symmetry in weight bearing.

Biofeedback instruments give more reliable, accurate and objective data on weight bearing than 'clinical examination' and can measure weight bearing during walking in contrast to 'scales'. However, such data can only be obtained when these instruments are calibrated and correctly applied to the human body. The costs of these devices, however, could limit the use in and outside physical therapy departments.

4.2. Research measurement techniques

The methodological quality of the *ambulatory devices* has only been extensively studied for the commercially available systems, of which the Pedar system has the greatest accuracy and repeatability. These studies, however, are restricted to short-term measurements for which these systems are mostly used. Data on the validity of the other ambulatory systems is limited to one article presented by the authors who developed the system. Measurement quality of the ambulatory devices depends on the type of transducers used. For instance, capacitive transducers have a fairly stable and linear response, but are relatively thick and less flexible compared to force sensing resistors (FSR). Another limitation of the capacity principle is the low sample frequency (100 Hz). More detailed descriptions of strain gauges, FSR, piezoresistive and piezoelectric transducers are given by Lord (1981), Cavanagh et al. (1992), Schaff (1993), and Cobb and Claremont (1995). The performance of the transducers can be influenced by the material used (e.g. wear and deformation) and by temperature changes, which can lead to e.g. hysteresis (Woodburn and Helliwell, 1996; Zhu et al., 1991; Cobb and Claremont, 1995), creep (McPoil et al., 1995; Chen and Bates, 2000) and temperature drift (McPoil et al., 1995; Woodburn and Helliwell, 1996; Miyazaki and Iwakura, 1978b; Zhu et al., 1991; Cavanagh et al., 1992) of the output signal. Especially when the device is placed inside the shoe the temperature can have an important effect on the sensitivity of the transducers (Cavanagh et al., 1992); however, only a few articles address the accuracy related to temperature range (King et al., 1972; Hennig et al., 1982). Although all systems can be calibrated, only a few systems can calibrate the sensors individually, and at a temperature related to the temperature in the shoe. The reported measurement range is 0–200 N/cm^2 and 0–250 kg, which is sufficient for weight bearing measurements. Only the early reports of Miyazaki report a relatively low measurement range, which may have been sufficient for their patients. Time resolution is 'good', with sample frequencies from 40–150 Hz which is sufficient for walking (Abu-Faraj et al., 1997; Cobb and Claremont, 1995; Cavanagh and Ulbecht, 1994).

The instruments can be semi-portable when using a cable, or portable using telemetry or a data logger to store data. These portable instruments can be used to measure weight bearing in the home situation during normal daily activities. Restrictions in collecting data over a longer time period are due to the software, energy consumption/power supply, and/or storage capacity.

Practicability (or simplicity in use) varies depending on whether the essential factors, e.g. attachment to the patient, calibration, data collection and analysis, are easy and not too time-consuming. Standard fixation of the portable devices on the subject or patient is with a belt on the hip. For patients, who have had hip surgery, this type of fixation is less comfortable and could even be painful. As the described systems are still relatively heavy and large for long-term daily measurements of weight bearing, future developments could focus on alternative forms of fixation. The commercially available instruments work with software packages, which simplify calibration and give extensive output information on e.g. (peak) force, step and stride length, and contact time. Weight of the device, location and type of sensors (barefoot, thickness of insole, slipper), and restriction due to cables determine the comfort for the patient and whether the system disrupts the gait pattern (Kljajic and Krajnik, 1987; Hermens et al., 1986). The ambulatory devices can be relatively expensive, especially when they require high sensor quality and calibration of the system.

Platforms have a high methodological quality to measure the ground reaction force and are therefore frequently used as a gold standard against which other systems, e.g. ambulatory devices, are evaluated. The transducers used in platforms are ideally suited for dynamic events, whereas for static measurements drift occurs over time. Quartz is used in the Kistler force transducers, and static measurements are more feasible with quartz than other piezoelectric material. Large forces can be measured for minutes and perhaps even hours; however low-level forces can be measured statically for much shorter intervals. This is why piezoelectric force transducers are often described as "quasi- or semi-static".

Standard platforms are restricted, compared to ambulatory devices, to single step measurements and therefore multiple trials are needed to gain reliable results. Other limitations are that the subject needs to hit the force plate correctly to obtain valid force data and problems of targeting may occur. To overcome the described limitations of standard platforms, long walkway platforms and treadmill force plates were developed. Another limitation for all platforms is their restriction to mostly a gait laboratory to measure the ground reaction force. To measure the ground reaction force in a rehabilitation environment, i.e. the clinic or home of the patient, an ambulatory device is the only option.

Specifically for measurement of ground reaction forces during (partial) weight bearing with walking aids one has to be aware that the patient does not place the aid on the force plate when collecting data of weight bearing on the foot (Mizrahi et al., 1985). Edwards (1986) concluded that for this kind of measurement a force shoe system is better than a force plate, because without a force shoe system it is difficult to simultaneously collect ipsilateral cane forces and extremity ground reaction forces because the proximity of the cane is too close to the affected foot and therefore distorts the force plate data. Platforms generally cost more than ambulatory systems, because of the extra equipment and personnel needed for correct placement in a laboratory. Furthermore, measurement costs are also higher because of the time-consuming measurements (Meggitt et al., 1981) and the trained personnel required for operating and calibrating these kind of systems (Fleming et al., 1997; Hall et al., 1996).

The choice of a technique for measuring weight bearing depends largely upon methodological, application, and feasibility aspects presented in this overview, but also on important aspects such as the clinical utilisation, the research question posed, the clinical set-up, and the available budget. To assess, for example, the amount of (partial) weight bearing after an orthopaedic procedure in the patient's home situation during one day, the portable *ambulatory device* technique seems to be the best option available. Commercially available ambulatory devices, however, still have limitations in collecting data over a longer (8–10 h) time period, and their relatively large size and weight can be strenuous for the patient. The new developments seen in the field of *ambulatory devices* technique are aimed at extending the measuring time, and at improved practicality for the researcher and clinician in data collection and data analysis. For these latter devices, however, mainly preliminary studies have been published about devices that are not (yet) commercially available.

This overview may support the selection of the most optimal technique to measure weight bearing for a specific application. Future development should focus on the limitations and disadvantages of the available techniques and instruments. Particularly for the *clinical examination* technique, information is lacking about its methodological quality. As this technique is often used in routine clinical evaluation of weight bearing, further research is needed on the validity and reliability of measuring weight bearing.

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