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Estimated ground reaction force in normal and pathological gait

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In clinical gait analysis, ground reaction force (GRF) is the gait parameter which can validate the state of disorder of the patient's movement. The purpose of this study was to explore the possibilities of employing the GRF derived from kinematics of the center of gravity (COG) in the study of dynamics of human gait. Gait data was collected for healthy able-bodied men and women and patients after ACL reconstruction who use larger lateral COG excursions during gait. Reasonable agreement between methods was found in fore-aft and vertical directions, where the methods differed by an average of less than 10% in either direction. Based on model predictions of the body's COG trajectory during walking, algorithms were developed to determine spatio-temporal gait parameters related to GRF characteristics. The suitability of calculating ground reaction forces using COG displacement in a patient population is questioned.

Key words: *motion analysis system, force platform, ground reaction force, center of gravity trajectory, human gait, ACL*

1. Introduction

The ground reaction force (GRF) characteristics during human walking can be an important descriptor of pathological gait [1]–[7] and can be easily obtained during routine clinical gait analysis as a complementary measure for standard data reporting [8], [9]. GRF measurement as a part of many professional and sophisticated computer systems, e.g. [10]–[13], was used earlier in the various studies, i.e. identification of muscle forces in human lower limbs during sagittal plane movements [14], [15] in the monitoring of the physiotherapy process after complex surgery or in hemiplegic, cerebral palsy and stroke patients [16]–[19].

The motion of the center of gravity (COG) during gait has long been known thanks to a number of kinematic studies, mainly by SAUNDERS & INMANN [9], [20], WINTER [1], [21], [22] and WHITTLE [23], [24]. Normal human walking is characterized by a periodic vertical displacement of the body COG that moves through a complete cycle of vertical motion with each step, or two cycles during each stride. The COG (often associated with trunk motion) rises twice during a gait cycle through a total of 4–5 cm, depending on gait speed [2], [9], [18], [25], [26], and is highest in the middle of the stance and swing phases, and lowest during double support. In the frontal plane, the COG moves from side to side once in a gait cycle being over each leg in stance phase. The total range of the sagittal movement is also 4–5 cm. As the body moves forward, the body COG's longitudinal speed varies, being highest during the double stance phases and lowest in the middle of the stance and swing phases [22], [27].

In recent decades, with technological advances, measuring the body COG as the center of a multi-segment human body can be the easy way to estimate the appropriate GRF characteristics. The method of retrieving GRF from kinematics of the COG is practical and feasible, but must first be compared with the well-established Newtonian computation of double-differentiating the COG trajectory. Plentiful studies have used

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the sacrum, pelvis, or trunk to estimate the position of the COG [19], [28]–[33]. This estimation assumes that the COG is fixed within the body, which may be a reasonable assumption in normal gait, but is imprecise in pathological gait [34]–[36]. A more sophisticated approach, the segmental analysis method, uses multiple markers to measure body segment positions, and incorporates an anthropometric model to calculate segmental center of mass positions for each recorded frame. These segmental centers of mass positions are then used to calculate the body COG [32], [34], [37], [38].

Although, at rough calculation, the motion of the sacrum and the segmental method are similar, they are not exactly the same – the COG is affected by the position of the lower and upper extremities which move independently of the trunk [39].

With the increased use of force platforms, COG displacement based on ground reaction forces during motions with continuous ground contact have been calculated through a simple dynamic equilibrium based on Newton's Second Law [40], [41]. The total force (associated with the resultant acceleration of the COG), GRF and body weight vectors can be expressed mathematically by the following equation:

$$\overrightarrow{M \cdot a} = \overrightarrow{GRF} + \overrightarrow{M \cdot g}\,, \tag{1}$$

where: M is the body mass, a – acceleration of the body center of mass, g – gravitational constant. Equation (1) states that the total force is the vector sum of ground reaction force and force of gravity. Direction of gravity is opposite to the laboratory axis so in the scalar notation for the vertical 'zet' component of the GRF vector, one can obtain:

$$GRF_z = Ma_z + Mg = M(a_z + g)\,, \tag{2}$$

or in the case of acceleration:

$$GRF_z[\mathrm{BW}] = \frac{a_z + g}{g}\,, \tag{3}$$

where GRF_z[BW] is the zet component of ground reaction force normalized to body weight. Finally, the loading rate of the ground reaction force in human gait is the degree of the slope of the GRF curve determined by the change of force (ΔF) divided by the time interval the change of force occurred.

Finally, the loading rate (LR) of the ground reaction force in human gait is the degree of the slope of the GRF curve determined by the change of force (ΔF) divided by the time interval at which the change of force occurred

$$LR = \frac{\Delta F}{\Delta t}\,. \tag{4}$$

The purpose of the present study was to check the possibilities of employing the ground reaction force derived from kinematics of the COG to the study of dynamics of human gait. This study aims to verify that the COG obtained during locomotion from a commonly-used model (Clauser's model) provides at least as reliable measurement of center of mass displacement as those obtained from the ground reaction force. Further goals were to determine whether the presence of gait pathology affected agreement between the two methods.

2. Material and method

2.1. Material

Total of seventy-six barefoot able-bodied adult volunteers participated in gait study while walking at their preferred speed. Of those fifty-three male (aged 31.5±9.7) and thirty-three female (aged 33.9±10.9) were patients of the Wrocław University College of Physiotherapy after the arthroscopic ACL reconstruction. Twenty-three healthy men (aged 22.1±3.2) and twenty-one women (aged 21.5±2.9) were classified into the control group.

All of the test patients underwent original physiotherapy process [42], [43] after the isolated ACL reconstruction. Following each stage of physiotherapy process, ACL-reconstructed patients were monitored by the motion analysis system. Stage 1 was held between 2–4 weeks postoperatively, stage 2: 5–8 weeks and stage 3: 9–12 weeks postoperatively.

Prior to participation, each subject signed a consent form approved by the ethical committee of the University School of Physical Education in Wrocław.

2.2. Method

A set of 18 reflective passive markers was used to denote the subjects' main upper and lower body parts as described by the Clauser model [44]. Additional four reflective markers were also placed on the force plate to register plate position and serve as the points of reference for transformation of local system of coordinates to global kinematic coordinates.

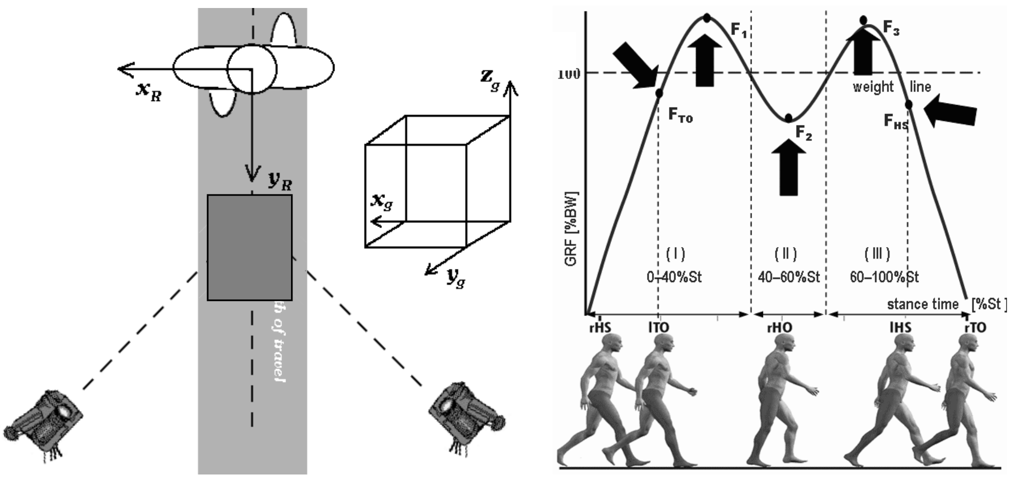
Kinematic data were recorded via a data acquisition system (Simi Reality Motion Systems GmbH, Unterschleissheim, Germany). Two 100 Hz digital JVC cameras were positioned ca. 4 m from the sagittal plane along the progression plane of the subject's gait path and were separated by an angle of approx. 80 deg (figure 1a).

The 6-degrees of freedom force plate (Kistler 9286A) equipped with strain gauges mounted underneath the four corners and the digital JVC camcorders were connected to the computer mainframe and synchronized with an optical starting signal. A cubic 1 m × 1 m × 1 m metal box was used for the calibration procedure and made up the laboratory frame of reference. Right-handed inertial reference system of coordinates was employed for both left and right body segments as well as the Global Coordinates System (GCS). The GCS is consistent with the Standardization and Terminology Committee of the ISB recommendations for standardization in the reporting of kinematic data [45] and is the same as the widely used Kistler force-platform convention with: X being the medio/lateral axis, Y – the anterior/posterior axis (the direction of travel) and Z – the proximal/distal (upward/downward) axis of coordinates [46].

acceleration of COG (a_z) was then computed by double differentiation operation while digitally processed (Butterworth 2nd order filter with the cut-off frequency of 6 Hz) [48].

The normal stance phase pattern of vertical forces generated by the foot and acting on the ground (figure 1b) has two peaks separated by a valley (often referred to as "M" shaped pattern). Commonly, the extremum events of vertical force are marked as $F1$, $F2$ and $F3$ [1], [2], [49]. The vertical force above the body-weight line (horizontal line indicating the body weight in figure 1b) represents action of upward acceleration of the body COG, and the force under the weight line represents the downward acceleration of the body COG.

The vertical component of the GRF vector was parameterized by the following indicators:



a) b)

Fig. 1. Measurement setup for the movement analysis (a) and search area definition against a background of a typical normal GRF pattern of stance phase (b). Two JVC digital cameras and Kistler force-plate are connected to a computer mainframe for the synchronized data acquisition. The calibration box, the right-handed subjects and general system of coordinates are also shown

Each subject began walking at a sufficient distance from the force plate so that the self-selected pace was attained prior to the foot of the test limb making contact with the center of the force plate. The length of the walkway (6 m) and the outer dimensions of Kistler platform surface limited the number of movement strides to around 3–5, depending on the subject's velocity. With the help of the SIMI Motion software, from the recorded motion it was possible to compute the time-related changes in location of each marker, and to divide each gait cycle into its characteristic phases: initial double stance, single stance, terminal double stance and swing phase for each limb.

Clauser's anthropometric data (the system built) were used to evaluate the path of body COG with the help of regression equations [41], [47]. The vertical

- the magnitude of force at the end of the initial double stance (F_{TO}), obtained as a force value at the toe-off time occurrence (t_{TO}) of the opposite limb;
- the maximal value of force (F_1) searched in the time range of 0–40% stance time (ST);
- time occurrence of F_1 (t_{F1}) and the time interval between initial contact ($t = 0$) and t_{F1} (Δt_{F1});
- the minimal value of force (F_2) searched in the single stance time range (40 to 60% ST);
- time occurrence of F_2 (t_{F2});
- the maximal value of force (F_3) searched in the time range of 60–100% ST;
- time occurrence of F_3 (t_{F3}) and the time interval between swing time (t_{sw}) and t_{F3} (Δt_{F3});
- the magnitude of force at the beginning of the terminal double stance (F_{HS}), obtained as a force value

at the heel-strike time occurrence (t_{HS}) of the opposite limb;

• force loading rate (*LR*) defined by equation (4) and measured at the time interval Δt_{F1};

• force unloading rate (*ULR*) defined by equation (4) and measured at the time interval Δt_{F3}.

All measured parameters in isolated cycles were averaged over 4 trials, and standard deviation was calculated. The *GRF* force was normalized with respect to body weight (BW) to objectively compare the results between subjects.

Normal distribution of data was checked by using the Shapiro–Wilk and Kolmogorov–Smirnov Normality tests. Equality of variances was checked by using either the *F*, Levene's or the Brown–Forsythe tests. Following physiotherapy process, the ACL-reconstructed subjects were treated as a dependent group. Two-way ANOVAs and a priori post hoc tests were used to determine differences between the two independent groups (ACL-reconstructed and control group) for measurement parameters for each of the four gait phases and five discrete points.

The analysis and data processing and evaluation were supported by the SIMI Motion analysis system (SIMI Reality Motion Systems GmbH, Unterschleissheim, Germany). All measurements were made in the Biomechanical Analysis Laboratory of the University of Physical Education in Wrocław (ISO quality standards: ISO 9001:2001).

3. Results

The analyses revealed a close correspondence between the direct (force platform) and indirect (film) method for measuring of the ground reaction force in tested parameters (table 1). No significant differences were observed between the results of men and women and for that reason only the results for the group of men are presented. No significant $p > 0.05$ differences in measured parameters were observed between tested groups, except the GRF value which at the beginning of terminal double stance phase (F_{HS}) appeared significantly smaller ($p = 0.05$) in the direct measurement method (table 1).

Moreover the *GRF* value, F_{TO} and F_{HS} were insignificantly larger in the film method (negative relative difference and *t*-value) mainly due to the relatively smaller values of force at the edge of *GRF* characteristics and due to an inaccuracy in determining the toe-off and heel-strike times in film method.

Table 1. Student's *t*-test significance of differences between direct (force platform) and indirect (film) methods. The relative difference between the mean values, *t*-value, degrees of freedom (*df*) and level of probability (*p*-value) for the *t*-test is also shown. Healthy men, $n = 23$

Indicators	Direct measurement mean ±*SD*	Indirect measurement mean ±*SD*	Relative difference	*t*-value	*df*	*p*-value
F_{TO} [BW]	0.80 ±0.05	0.81 ±0.11	–1.2%	–1.25	22	0.09
F_1 [BW]	1.14 ± 0.09	1.17 ±0.12	–2.5%	–1.40	22	0.18
t_{F1} [s]	0.15 ±0.05	0.15 ±0.09	0.0%	0.63	22	0.54
LR [BW\s]	7.6 ±0.8	7.8 ±1.3	–2.5%	–8.76	22	0.10
F_2 [BW]	0.88 ±0.11	0.87 ±0.15	1.1%	1.25	22	0.22
F_3 [BW]	1.17 ±0.10	1.18 ±0.11	–0.8%	–0.20	22	0.84
T_{Sw}–t_{F3} [s]	0.17 ±0.02	0.17 ±0.03	0.0%	0.05	22	0.96
ULR [BW\s]	6.9 ±1.3	7.3 ±1.6	5.5%	0.48	22	0.63
F_{HS} [BW]	0.80 ±0.05 *	0.82 ±0.13 *	–2.4%	–2.88	22	0.05

* Significance level $p < 0.05$.

The measured parameters (by the indirect film method) positively diversify the change in the magnitude of *GRF* in time of physiotherapy process in ACL-reconstructed group (table 2). The parameters F_1, F_2 and F_3 followed by *LR* and *ULR* demonstrated the most significant change of all measured parameters. The *GRF* value at the end of initial double stance (F_{TO}) increased significantly (ca. 16%) between stages one and two. It increased insignificantly between the second and third measurements, just around 1%. Similar relations were found in the *GRF* value at the beginning of terminal double stance (F_{HS}). Force at heel-strike increased significantly (ca. 21%) between stages one and two and insignificantly (ca. 1%) between stages two and three. The first maximum of *GRF*(F_1) increased considerably (on average of 18%) between stages one and two, and subsequently 9% between stages two and three. The local minimum of *GRF* in the middle of single stance (F_2) was decreasing during the physiotherapy process. Between stages one and two it significantly decreased (ca. 7%), and

Table 2. Results of the two-way ANOVAs and a priori post-hoc tests for the ACL-reconstructed and control groups in the indirect film method

Parameters	ACL-reconstructed group (N = 86)			Control group (N = 75)
	Stage 1 (2–4 t.)	Stage 2 (5–8 t.)	Stage 3 (9–12 t.)	
F_{TO} [BW]	**0.70 ±0.06***	**0.81 ±0.08**	**0.82 ±0.08**	**0.82 ±0.05**
Men	0.71 ±0.06*	0.81 ±0.08	0.82 ±0.08	0.82 ±0.05
Women	0.70 ±0.06*	0.82 ±0.07	0.82 ±0.07	0.82 ±0.05
F_1 [BW]	**0.91 ±0.06***	**1.07 ±0.09***	**1.19 ±0.11**	**1.17 ±0.02**
Men	0.91 ±0.06*	1.08 ±0.09*	1.20 ±0.12	1.17 ±0.02
Women	0.92 ±0.06*	1.05 ±0.08*	1.17 ±0.08	1.16 ±0.02
Δt_{F1} [s]	**0.11 ±0.07**	**0.13 ±0.01****	**0.15 ±0.04**	**0.15 ±0.00**
Men	0.11 ±0.07	0.13 ±0.02**	0.15 ±0.03	0.15 ±0.00
Women	0.11 ±0.08	0.13 ±0.01*	0.14 ±0.05	0.15 ±0.01
(LR) gradient F_1 [BW\s]	**8.5 ±0.1***	**8.2 ±0.6**	**8.1 ±0.3**	**8.0 ±0.1**
Men	8.5 ±0.1*	8.1 ±0.6	8.0 ±0.2	8.0 ±0.1
Women	8.4 ±0.2**	8.2 ±0.5*	8.1 ±0.3	8.0 ±0.0
F_2 [BW]	**0.87 ±0.07***	**0.81 ±0.06***	**0.80 ±0.18**	**0.81 ±0.05**
Men	0.86 ±0.06*	0.79 ±0.06*	0.78 ±0.18	0.81 ±0.05
Women	0.88 ±0.09*	0.84 ±0.06*	0.81 ±0.19	0.80 ±0.05
F_3 [BW]	**0.90 ±0.06***	**1.05 ±0.09***	**1.17 ±0.12**	**1.11 ±0.03**
Men	0.90 ±0.06*	1.06 ±0.09*	1.19 ±0.12	1.11 ±0.03
Women	0.91 ±0.07*	1.05 ±0.08*	1.14 ±0.13	1.12 ±0.02
Δt_{F3} [s]	**0.18 ±0.03***	**0.20 ±0.03***	**0.17 ±0.02**	**0.16 ±0.01**
Men	0.17 ±0.03*	0.20 ±0.03*	0.17 ±0.02	0.16 ±0.01
Women	0.18 ±0.04*	0.21 ±0.03*	0.18 ±0.03	0.16 ±0.02
(ULR) gradient F_3 [BW\s]	**5.2 ±0.2**	**5.3 ±0.5***	**6.8 ±0.3**	**7.0 ±0.1**
Men	5.2 ±0.2	5.3 ±0.6*	6.8 ±0.3	7.0 ±0.1
Women	5.1 ±0.3	5.2 ±0.5*	6.7 ±0.4	7.0 ±0.2
F_{HS} [BW]	**0.67 ±0.10***	**0.81 ±0.09**	**0.80 ±0.05**	**0.82 ±0.09**
Men	0.67 ±0.09*	0.82 ±0.09**	0.80 ±0.05	0.82 ±0.08
Women	0.67 ±0.15*	0.80 ±0.08	0.81 ±0.07	0.81 ±0.09

* Significance level $p < 0.1$.
** Significance level $p < 0.05$.

between stages two and three it decreased less than around 1%. At first, the second maximum of *GRF* (F_3) increased considerably (ca. 17%), then increased significantly (ca. 11%) between subsequent measurements. The loading rate (*LR*) decreased significantly between stages one and two (ca. 5%) and decreased insignificantly between stages two and three. The unloading rate (ULR) at the end of stance slightly increased at first (2%) and then it significantly increased (more than 28%) between stages two and three.

No statistically significant differences were noted between the last stage of ACL-reconstructed group and the control group in all tested parameters. Moreover, the result of measurement of the first and second maxima of *GRF* characteristics (F_1 and F_3) for the third measurement of ACL-reconstructed group was more than 5% higher than for the control group and the minimum (F_2) less than 5% lower. Those results, however, were not statistically significant.

4. Discussion and conclusions

The three-dimensional kinematic gait analysis was conducted on normal and ACL-deficient subjects to test the usefulness of ground reaction force (*GRF*) measurement obtained from the kinematic data of the body center of gravity (COG) in clinical condition.

The indirect film method used to calculate GRF from the double differentiation of the COG trajectory compared reasonably well with the direct measurement. The film method was shown to be prone to errors from numerical operations (mainly filtering and differentiation). The values of excursion differences were comparable with the results reported by EAMES et al. [32], who also determined higher excursions from the *GRF* integration, and with the results of SAINI et al. [34] and THIRUNARAYAN et al. [50], who only considered vertical displacements and who used simplified upper body model containing only one head–arms–trunk (HAT) segment. The two methods showed deeper disagreement in the current study than in that by GARD et al. [51], who used different, more sophisticated anthropomorphic model together with six force platforms and found broad agreement between of the vertical *GRF* characteristics these two methods. The anthropomorphic model used in this study (the Clauser model) was of particular interest because it is commonly used in both clinical and research settings. The agreement between methods in the vertical direction was diminished by the incidence of pathology. One of the fundamental parameters of *GRF* characteristics is the maximal value of force at the beginning of single stance (F_1) and the loading rate (*LR*). SCHAFFNER et al. [52], while testing the gait of astronauts during locomotion in weightlessness (0G) and under normal (1G) conditions, received F_{max} = 1.17 BW and *LR* = 7.29 BW\s. McCRORY et al. [53] obtained slightly lower results: 1.02 BW and 5.22 BW\s, respectively. The above results seem to confirm the findings of STACOFF's et al. [54] (F_{max} = 1.19 BW and *LR* = 7.92 BW\s). Contrary to what was expected, the loading rate proved to be the highest in the affected limb in the ACL group and the lowest in the normal group, probably because of shorter support time for the affected limb. The last conclusion may be related to the results of BUS [55], who found higher loading rates in older-aged runners as compared with younger-aged runners, and attributed this difference to a loss of shock-absorbing capacity in the older runners.

Age of the test subjects was found to be a factor which should be considered, because the young age group walked faster and produced larger vertical *GRF* maxima during level walking than the middle and old age groups.

To conclude, the indirect film method slightly overestimates the computation of ground reaction force in several measured parameters (table 1) but meets the most urgent physiotherapeutic needs and with a rough approximation can replace the direct measurement method in a clinic unit where no forceplate is present or the direct measurement not viable. Therefore, the film method of acquiring *GRF* characteristics and considering the relative values only can be a good indicator of the progress in physiotherapy. Calculating the *GRF* from a kinematic of COG rather than from the force platforms, other than being sensitive to errors in anthropometry, marker placement, and modelling, requires a complex 3-D motion analysis system. It does, however, have important advantages. Using kinematics to calculate the COG allows accurate calculation of its position, not just displacement, in both globally and subject-fixed frames of reference [37]. What is more, the accuracy of the indirect method does not depend on a steady, symmetric gait. Simultaneous comparison of kinematics and COG movement also makes it possible to measure how particular kinematics affects the COG movement. The indirect method assumes segment masses as fixed proportions of body mass and is therefore not dependent on body mass estimation. The indirect film method can also be used to measure COG during flight activities (no ground contact). The suitability of calculating ground reaction forces using COG displacement in a patient population is uncertain. On one hand, the kinematic centroid in pathologic gait is more susceptible to errors in segment parameters and marker placement (body deformity), on the other hand, it allows us to arrive at reliable results that are at least within the range of doubt of the COG differentiation, and are more valuable when interpreting patients' 3-D gait data.

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