



Original article

Gait-force model and inertial measurement unit-based measurements: A new approach for gait analysis and balance monitoring

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Abstract

Background/Objective: This work describes a new approach for gait analysis and balance measurement. It uses an inertial measurement unit (IMU) that can either be embedded inside a dynamically unstable platform for balance measurement or mounted on the lower back of a human participant for gait analysis.

Methods and Results: The acceleration data along three Cartesian coordinates is analyzed by the gait-force model to extract bio-mechanics information in both the dynamic state as in the gait analyzer and the steady state as in the balance scale. For the gait analyzer, the simple, noninvasive and versatile approach makes it appealing to a broad range of applications in clinical diagnosis, rehabilitation monitoring, athletic training, sport-apparel design, and many other areas. For the balance scale, it provides a portable platform to measure the postural deviation and the balance index under visual or vestibular sensory input conditions. Despite its simple construction and operation, excellent agreement has been demonstrated between its performance and the high-cost commercial balance unit over a wide dynamic range.

Conclusion: The portable balance scale is an ideal tool for routine monitoring of balance index, fall-risk assessment, and other balance-related health issues for both clinical and household use.

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Introduction

An inertial measurement unit (IMU) is a device that measures the acceleration vectors. Typically, it consists of three identical units packaged along orthogonal orientations in the Cartesian-coordinate plane. Over the last decade, this technology has undergone explosive growth in both size and cost reduction, as well as performance improvement in sensitivity, dynamic range, and data rate. These changes transformed IMUs from a cumbersome scientific instrument with narrow and specific applications to a low-cost and high-performance tool that can be used in many areas. One important area is the monitoring of human movement^{1,2} both in dynamic gait activity or passive activity, such as maintaining balance. This work demonstrated its applicability as a simple and effective

way to diagnose physical gait impairment and to assess the effectiveness of therapeutic treatment that can lead to better personal health and quality of life.

Gait analysis is the study of human movements, such as walking, running, and other forms of physical activities. The earliest form of analysis appeared more than a century ago using cinematographic techniques to capture sequential images of a person in motion. Results were analyzed by recording the displacement change of a specific part of the body meticulously from frame to frame. Fueled by new technologies, the approach has evolved with the addition of additional tracking markers, higher image-capture rates, and in some cases the addition of gyroscopes to measure angular changes. However, despite the constant stream of new implementations, the fundamental core approach remains the same, and there are several issues yet to be resolved.

The necessity to extract useful information from recording and analyzing a large data volume with complex geometric

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transformation algorithms has limited the clinical value of traditional gait approaches. However, aside from this limitation, there are several other issues. One of the key issues is its elaborate setting in a confined laboratory environment that not only limits the type of physical activities to be studied, but also hampers gait measurement under a natural stance. In order to expand the types of physical activities, such as long-distance running and ascending/descending stairs, a larger and less restrictive setting is preferred. On a more fundamental side, another issue is the relationship between the measured quantities in the time-space domain and biomechanical models used to describe gait in the force-time space. Transforming displacement data to force vectors requires second-derivative calculus operations that can eliminate details due to insufficient resolution. The addition of optical markers and cameras to track additional parts of the body aims to solve this problem, but simultaneously adds complexity. Gait analysis with a single IMU described in this work provides an effective solution to both problems.

The measurement of body balance is another area where IMUs can excel. The traditional approach index measurement is performed by using a set of force plates embedded in a static platform on which the human participant stands for testing. The vertical projection of the center of gravity (COG) of the individual is determined by triangulating the force-plate sensor inputs. This approach only senses the vertical force, whereas forces in lateral directions propelling side movements are only indirectly inferred. Here, the balance index was determined by an IMU that tracks movement along all three directions in real time.

Methods and Results

Single-point gait analysis

In single-point gait analysis, an IMU is mounted on the lower back in the proximity of the anatomical COG of the individual being tested. At this location, the counterforce generated from ground contact during each heel strike is mostly damped by the ankle, knee, hip and other joints in the lower extremity of the human participant (Fig. 1).

Therefore, under an ordinary walking stance, the acceleration along three axes measured at this location can be approximated as the acceleration of an individual's COG. For mild physical activity, such as walking, 100-Hz data has been shown to be adequate. Another important approximation not accounted for is the rotational movements along three axes (pitch, roll, and yaw). Under normal gaits, this is a valid approximation. Even the inclusion of rotational variation will add only some analytical complexity, whereas the basic core algorithm remains unchanged.

A typical set of raw data shown in Fig. 2 depicts three groups of measurements representing the acceleration along three axes. Instead of using anatomical terms, such as anterior/posterior, etc., physics-based terms surge (forward/back), sway (left/right), and heave (up/down) are used for simplicity. Because acceleration is related to force by the mass of the

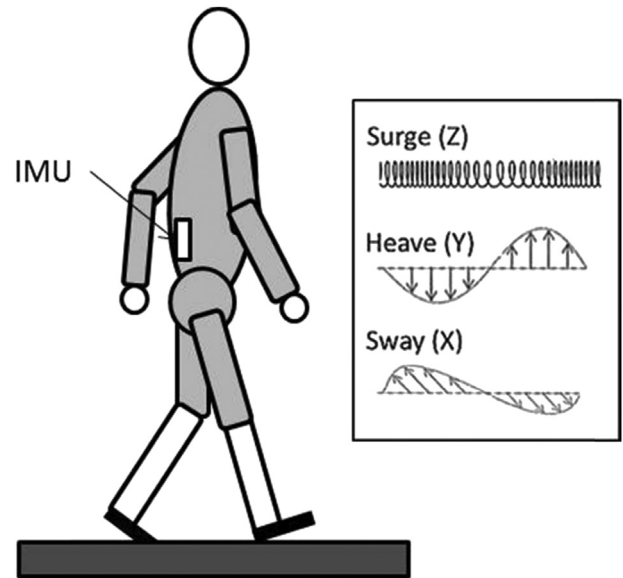


Figure 1. Configuration of gait analysis with a single inertial measurement unit.

participant, such data can be regarded as the normalized force. Therefore, this experimental setting yields information about the magnitude of force in different directions, which is more closely related to the biomechanics of the gait as compared with displacement measured by the traditional gait approach.

Displacement can be calculated from the acceleration value by a double integration with the appropriate boundary conditions. In the current measurement setting, given the IMU sensitivity at $\sim 0.003\text{ g}$ ($g = 9.8\text{ m/s}^2$ is the gravitational acceleration constant), the displacement resolution is only 0.05 mm, a sensitivity far exceeding the limit of image-capture measurements, despite the simplicity of the measurement.

As the data in Fig. 2 and the inset in Fig. 1 indicate, the gait pattern can be represented by superimposing three oscillatory movements in surge, sway, and heave. Heave is the only movement that makes contact with the ground and produces counterforces that can be regarded as the energy intake to fuel the gait. Surge is the mechanism that converts the intake

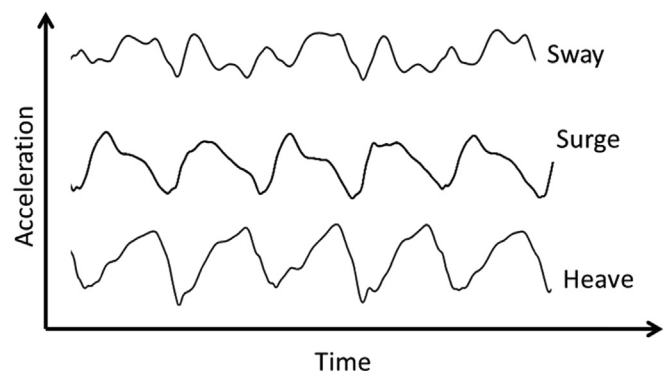


Figure 2. Raw data from inertial measurement unit-based gait analysis.

energy from Heave to propel the motion forward. Sway motion from side to side is primarily responsible for keeping the body in balance. The data-analysis model is based on treating the body movement during each stride as a whole system in motion. The energy petition and timing relationship among different directions determine the gait quality, and any deviation can be regarded as a gait flaw and abnormality. There were several reports in the past regarding the use of IMU for gait analysis.^{3,4} Here, we described a new development in IMU-based gait analysis. This approach is completely different from those described in previous reports in data collection and analysis. The core model is based on a new concept known as gait force,^{5,6} which is used to extract vital gait information. Key components are described in the following sections.

Gait-event designation

Each stride consists of a series of key events starting with the heel strike of one foot, followed by the stance phase, the heel strike of the other foot and the swing phase, and terminating with the heel strike of the same foot as shown in Fig. 3.

On a finer scale, this cycle can be described by a subset of events denoting movements involving the knees and toes. Identifying the precise time of the heel strike is the most essential task that anchors further gait analysis. Conventional gait analysis relies on frame-to-frame analysis of the sequential images to pinpoint the heel strike. This process is painstaking, with uncertainty determined by the image-capture rate. However, in single-point gait analysis with an IMU sensor, heel strike can be precisely determined with accuracy within 0.01 sec. by analyzing the velocity of vertical motion. Another key event is the instant just prior to toe off or at the end of the stance phase when both feet are flat and making contact with ground. The precise time of this event can be determined from the height of the COG. We have applied this algorithm to analyze over 100 gait samples and obtained the ratio of the

stance phase to the entire cycle in a narrow range from 62% to 65%, which was consistent with accepted values. The designation was also consistent with gait-force spectrum analysis to be described in later sections.

Gait force

We propose a new concept (gait force) to describe the trajectory of the acceleration vector involving the sum of all components in a reference frame, with the COG as its origin. For a series of strides, the trajectory can be depicted as a three-dimensional (3D) image known as a gait-force image. Every individual has a unique gait-force image, similar to the one shown in Fig. 4A. For a clearer view and further analysis, the 3D image is decomposed into three separate two-dimensional (2D) images shown in Fig. 4B by collecting each as a slice of the 3D figure at different coordinates and orientations.

The image labeled Sway/Surge describes the movement in acceleration space from the side, the image labeled Sway/Heave describes the view from the rear, and the image labeled Surge/Heave describes the view from the top. The shape and symmetry of these images contains vital information about individual gait quality. The gait-force imaging can be regarded as a new medical-imaging technique and a diagnostic tool for rapid viewing and assessment.

Gait-force spectrum

Gait-force spectrum (or gait-power spectrum) is another new concept that represents the information extracted from the raw data in the form of a power spectrum in the time domain and consists of repetitive groups of peaks from stride to stride. The vertical axis represents the power level associated with the inflow and outflow of energy. A poor gait form is associated with frequent small and sudden movements that appear as closely spaced sharp peaks. The opposite occurs in the

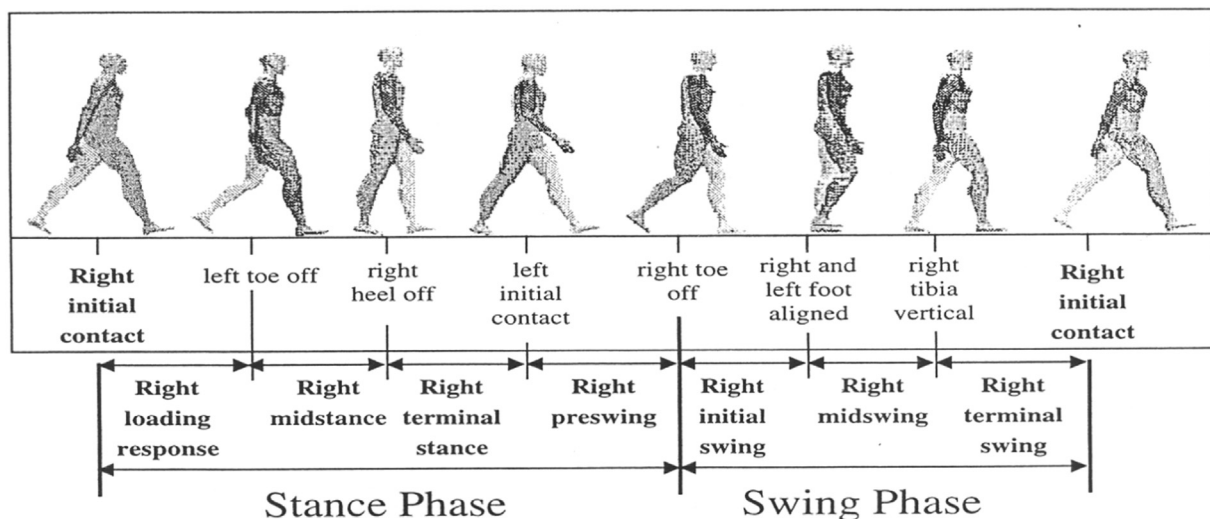


Figure 3. Phases of a gait cycle.

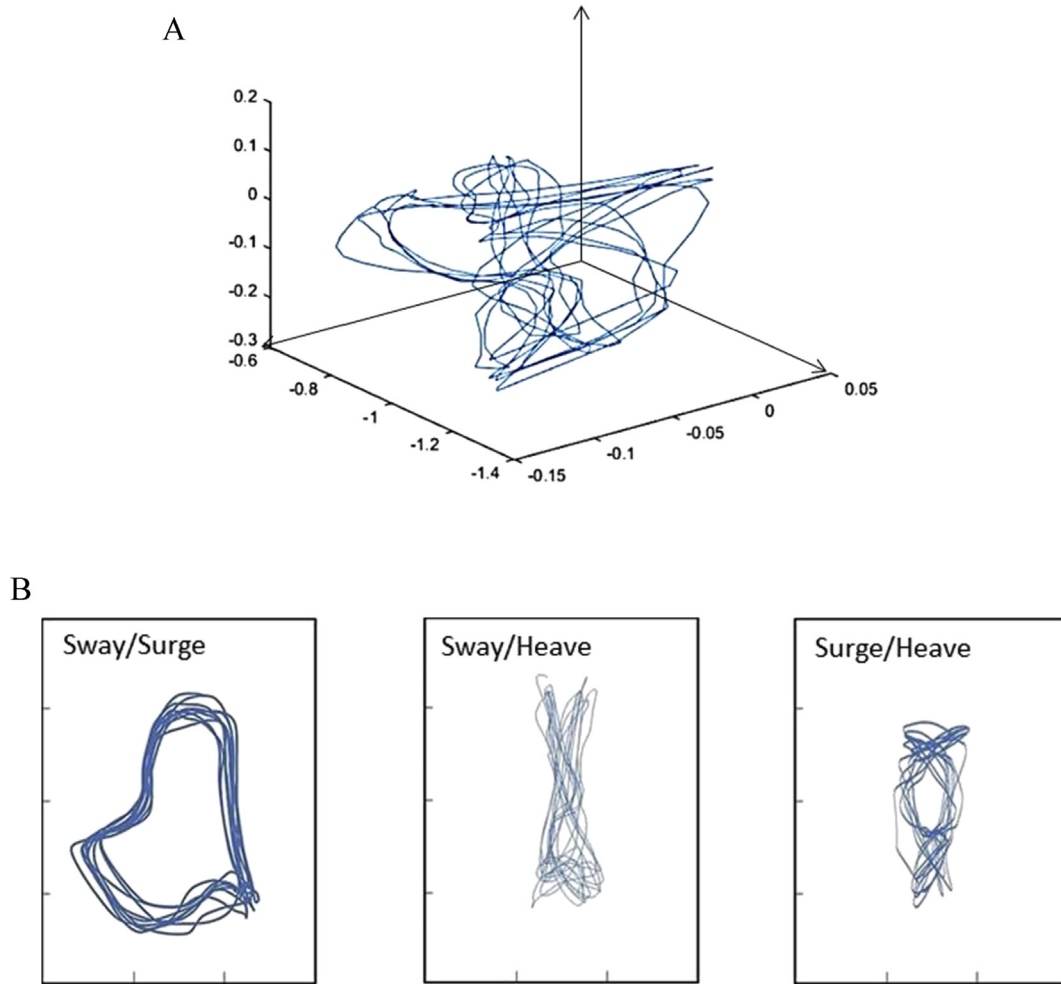


Figure 4. (A) Gait-force image and (B) two-dimensional gait-force image components.

presence of healthy gait forms. Fig. 5 shows an example of the raw data and the corresponding gait-force spectrum covering five steps.

This gait pattern shows two places for energy intake during each step. One intake occurs shortly after heel strike due to knee bending and extension. This triggers the surge motion forward after a slight delay, which is characteristic for each individual. The second energy intake is powered by toe-push. The consecutive energy injection in a timely manner is the key to an efficient gait and ensures sustainable energy to power the motion forward. This is reflected in the surge motion in the raw data.

Due to its high sensitivity, gait-force spectrum can be used to detect small gait flaws. Some examples in diagnosing leg-length discrepancy (LLD) conditions and flat-foot syndrome (FFS) are briefly described here. LLD is a clinical anatomic condition that affects ~80% of the population, and should be identified at as early an age as possible for proper treatment. If untreated, this condition will eventually lead to the formation an S-shaped spine and/or other diseases. Our preliminary studies showed that LLD can be identified with a high degree of reliability using this technique (Fig. 6). The characteristic LLD features included alternating strength in the acceleration

in one or more directions example insuch as the heave motion shown. Secondary features included the lack of knee 3bending of the short leg and the propagation of small amounts of undamped counterforces following heel strike. All symptoms

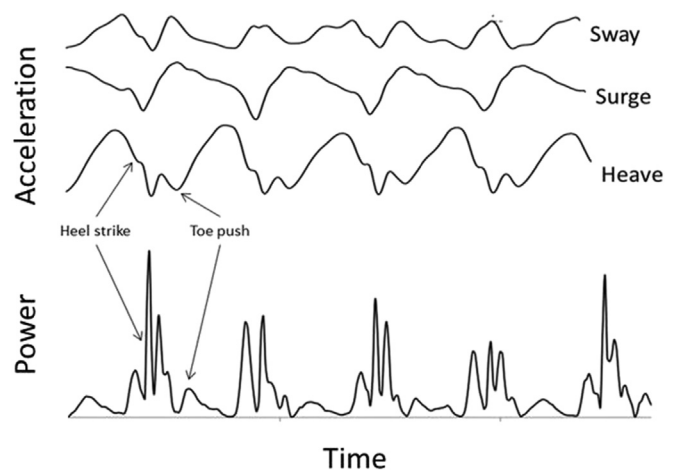


Figure 5. Raw data and the corresponding gait-force spectrum involving five steps.

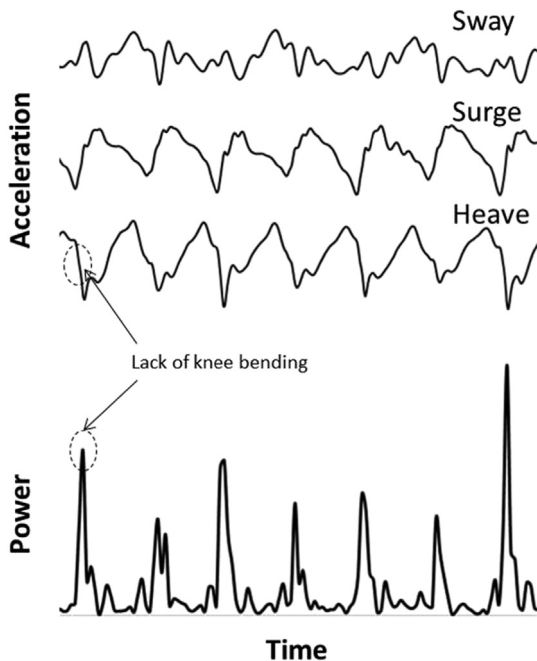


Figure 6. The gait measurement of a person with leg-length discrepancy.

were reflected as characteristic feature in the raw data, as well as the gait-force image discussed in the previous section.

Flat Foot Syndrome or FFS is another pedorthic condition that affects >60% of the population. The lack of sufficient foot arch fails to produce the necessary leverage for toe push; therefore, the energy that propels forward motion predominately comes from the knee action after initial heel strike. Without replenishing energy by toe push, surge-motion acceleration diminishes rapidly without any sustainability (Fig. 7). The situation can be improved with proper corrective footwear.

Other clinical examples

We also measured the gait force of people with special medical conditions. Fig. 8 shows the gait of a patient with

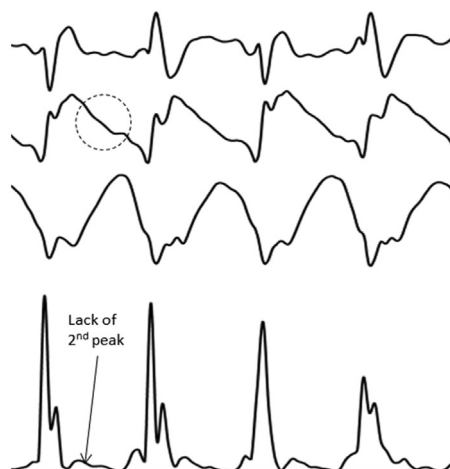


Figure 7. Gait pattern of a patient with flat-foot syndrome. The dashed circle indicates lack of sustainability in surge motion due to the absence of toe push.

Parkinson's disease. The characteristic "frozen gait" symptom is clearly visible.

Fig. 9 shows the gait of a patient during rehabilitation recovery following hip replacement. The pattern shows asymmetry between the left and right steps, as well as very weak peaks (~30% of normal gait) in the gait-force spectrum and an erratic heave pattern. These results suggested the usefulness of this technology as a tool to monitor rehabilitation progress.

Other applications

In addition to medical diagnosis, the versatility and simplicity of this technology enables it to be used for other applications. An important area is the quantitative assessment of the effectiveness of therapeutic solutions, such as pedorthic insoles and special footwear such as prosthetic for patients suffering gait-impairment conditions. Traditional assessment is based on patient feedback and practitioner experience. However, the lack of a quantitative measurement can lead to uncertainty in making the best decisions. This tool allows unambiguous assessment of therapeutic effectiveness and identification of areas requiring further improvement. Fig. 10 shows a set of gait-force spectra measured for an individual wearing shoes with or without therapeutic insoles. Peaks in a gait-force spectrum represent energy flow between the individual and the external environment. Poor gait can lead to spiky and frequent energy flow in and out of the system and manifests as characteristic sharp and closely spaced peaks in the gait-force spectrum. The sharpness and closeness of these peaks are caused by discomfort during the gait cycle as the body is constantly shifting in search of a more comfortable form. Based on this interpretation, it is clear that the insole has positively improved the gait of this individual.

Other applications include selection of the most suitable events for athletes to pursue and more effective athletic training through identification of areas requiring improvement. This simple gait-analysis approach can also be applied to study young children when the collection of normal gait data



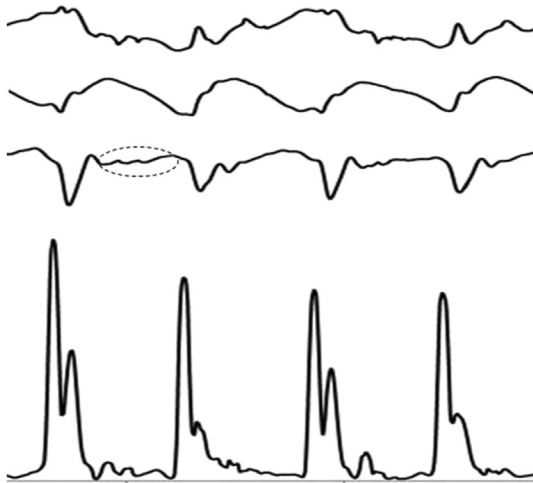


Figure 8. Gait-force pattern and gait-force image of a patient with Parkinson's disease. The circled flat region between heel strikes corresponds to "frozen gait".

is virtually impossible. Finally, this approach is not limited to human participants, and can also be used to monitor the gait of animals, which is impractical using the convention approach.

Study of body balance

Complex neuromuscular coordination is vital to maintaining balance. Traditionally, the ability to perform this task is quantified by a balance index that is measured with large equipment used exclusively by medical professionals. The bulky size and high-cost prohibit such equipment from becoming a household health-monitoring tool for the general public. In addition, most balance-monitoring tools measure the projection of an individual COG on a static platform with embedded force-plate sensors, which accounts for only the downward force, with contributions from lateral movements to shift COG inferred indirectly. We have developed an IMU-

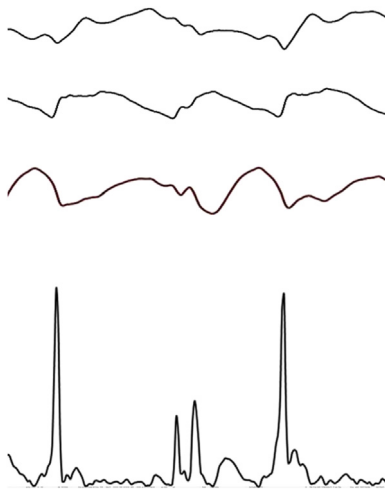


Figure 9. Raw gait data and gait-force spectrum of a patient recovering from hip-replacement surgery.

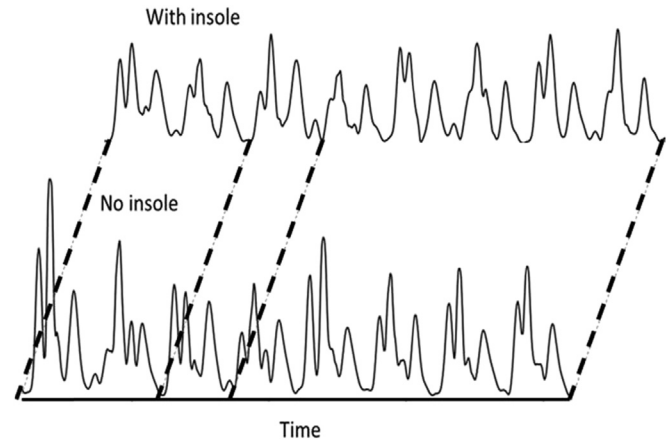


Figure 10. Gait-force spectrum of an individual walking with and without wearing pedorthic insoles.

based portable balance scale to alleviate this limitation.⁷ In this equipment, an IMU is integrated with a spring-loaded dynamic platform on which the human participant stands. Tests can be performed with the eyes of the participant either open or closed to separately study the effects of visual sensory input and vestibular sensory input in under 20 sec. Output of the IMU can be represented as the trajectory of the COG during measurement (Fig. 11).

From this data, useful information can be extracted, including the balance index, postural deviation, and fall-risk assessment, as well as the most likely direction of falling. Balance-index measurements were compared with those made with the Biodex SM Balance under identical conditions for comparison. Results in Fig. 12 show excellent agreement over a wide dynamic range, thereby validating the performance of this approach.

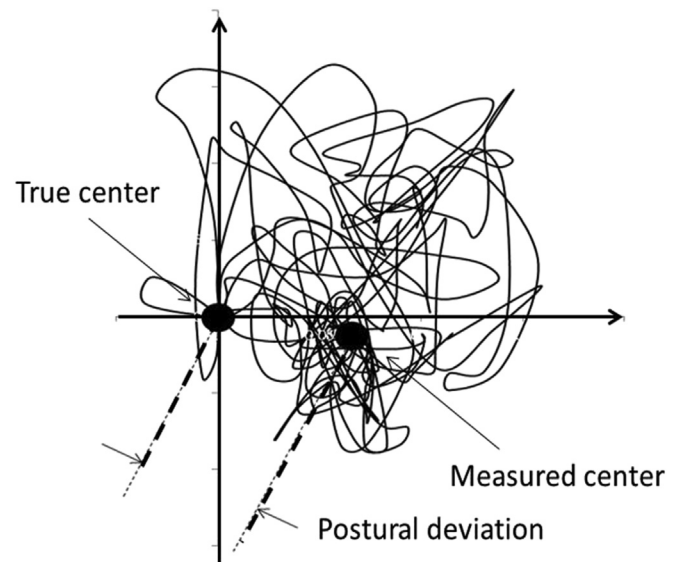


Figure 11. The trajectory of center of gravity during balance. The solid circles represent the center of the unloaded platform and the center of mass of the acceleration vectors during balance measurement. Their difference represents a measurement of postural deviation.

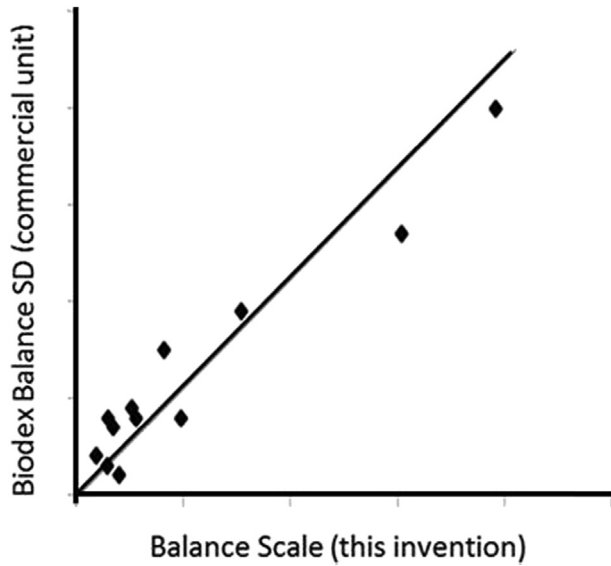


Figure 12. Calibration of the device presented in this work versus that of Biodex Balance SD, a commercial balance-testing apparatus.

The ability of an individual to maintain balance decreases with age. For the same individual, balance maintained using only vestibular sensory input is inferior to that using visual sensory input. We have observed that the ability to maintain

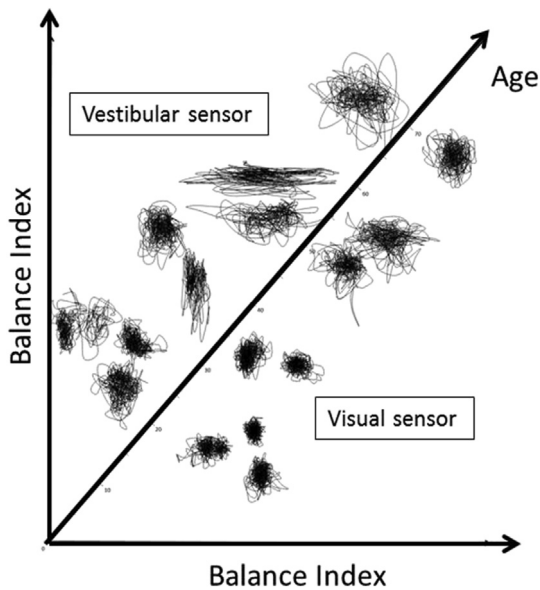


Figure 13. The dependence of balance index on age and sensory input.

balance with just vestibular sensory input starts to deteriorate at ~30 years of age. A summary of these effects is shown in Fig. 13.

Additional information that can be extracted is the most likely direction to fall in the case of losing balance. This is measured as the density of the acceleration vectors in the four quadrants, with the quadrant having the highest density representing the most likely direct to fall.

Conclusion

This work described a new approach for gait analysis and balance measurement with a single IMU and data analysis based on a gait-force model. This method reduced the complexity associated with traditional methods and proved effective at yielding vital information without sacrificing quality or depth. The simplicity and portability of this method enable a broad range of new applications.

Conflicts of interest

The authors have no conflicts of interest to declare.

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