Use of Accelerometry to Classify Activity Beneficial to Bone in Premenopausal Women

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[1]Department of Sport and Health Sciences, College of Life and Environmental Sciences, University of Exeter, Exeter, England, UNITED KINGDOM; [2]Exercise, Health, and Human Performance Group, Sansom Institute for Health Research, School of Health Sciences, University of South Australia, Adelaide, SA, AUSTRALIA; and [3]Health and Use of Time Group, Sansom Institute for Health Research, School of Health Sciences, University of South Australia, Adelaide, SA, AUSTRALIA

ABSTRACT

STILES, V. H., P. J. GRIEW, and A. V. ROWLANDS. Use of Accelerometry to Classify Activity Beneficial to Bone in Premenopausal Women. *Med. Sci. Sports Exerc.*, Vol. 45, No. 12, pp. 2353–2361, 2013. **Purpose**: The aims of this study were to quantify the relation between ground reaction force (GRF) and peak acceleration from hip- and wrist-worn accelerometers and determine peak acceleration cut-points associated with a loading rate previously demonstrated as beneficial to bone (43 body weights (BW)·s^{-1}) in premenopausal women. **Methods**: Forty-seven premenopausal women (age, 39.2 ± 5.6 yr; mass, 65.9 ± 11.0 kg; height, 1.67 ± 0.06 m) performed walking (slow, fast, and with bag), floor sweeping, running (slow and fast), jumping (low, <5 cm; high, >5 cm), and box drop (20 cm) trials. Peak accelerations were sampled at 100 Hz by GENEActiv and ActiGraph GT3X+ accelerometers (ActiGraph LLC, Pensacola, FL) worn at the hip (vertical and resultant) and the wrist (resultant). A force plate (960 Hz, AMTI) was used to assess peak vertical GRF and peak loading rate for eight steps per activity. Receiver operating characteristic curves were used to determine the optimal peak acceleration cut-points associated with a loading rate of 43 BW·s^{-1} in 37 participants, and these cut-points were cross-validated in the remaining 10 participants. **Results**: For all activities combined, peak accelerations were positively and significantly ($P < 0.001$) correlated with peak vertical GRF (hip $r > 0.8$, wrist $r > 0.7$) and peak loading rate (hip $r > 0.7$, wrist $r > 0.57$). Irrespective of monitor type and wear site, peak acceleration discriminated between loading rates above and below 43 BW·s^{-1} with high levels of accuracy (area under the curve >0.92, $P < 0.001$). Overall classification agreement was >85% for both monitors worn at either the wrist or hip in the cross-validation sample. **Conclusion**: GENEActiv and ActiGraph GT3X+ accelerometers worn at the wrist or hip can be used as an unobtrusive tool to identify the occurrence of loading rates likely beneficial to bone in premenopausal women during their daily activity.
Key Words: ACTIVITY MONITORS, GENEACTIV, ACTIGRAPH, LOADING RATE, PHYSICAL ACTIVITY

Although bone-preserving action is important across the lifespan, the premenopausal phase is a critical window when women have the scope to optimize bone accrual and maintain or increase bone density to minimize bone loss during the postmenopausal years (19). Physical inactivity, in addition to nutritional, environmental, and genetic factors, is an established risk factor for osteoporosis (1,19). Specifically, short bursts of dynamic activity characterized by high-impact forces and loading rates have been found to increase bone mineral density (BMD) in premenopausal women (2,14,34). Until recently, it has not been possible to measure activity with the resolution necessary to capture this type of activity pattern in free-living individuals. However, advances in commercially available triaxial accelerometry-based activity monitors (e.g., GENEActiv (ActivInsights Ltd, Cambridgeshire, UK) and ActiGraph GT3X+ (ActiGraph LLC, Pensacola, FL)) allow the collection and storage of raw acceleration data at frequencies of up to 100 Hz for up to 7 d at a time. Both monitors can be worn at the hip but are primarily designed to be worn at the wrist, which leads to greater participant wear time (33). To use these monitors to assess activity beneficial to bone, the raw acceleration output from accelerometers worn at both the hip and the wrist needs to be calibrated with mechanical measures relevant to bone health, e.g., ground reaction force (GRF).

Quantifying external force on and estimating internal force in the body are important for understanding the influence of loading on bone geometry, strength, and mass (18). In animal research, bone strain magnitudes have been found to be linearly proportional to the magnitude of the externally applied load (16). The use of external measures of load (e.g., peak vertical GRF) to provide an indirect estimate of strain magnitude is therefore justified. Peak vertical GRF has been found to correlate strongly with peak accelerations from hip- and wrist-worn activity monitors in young adults (28), and algorithms have been developed to estimate peak vertical GRF on the basis of accelerations from hip-worn activity monitors in girls and boys during walking and running (24).

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Submitted for publication March 2013.
Accepted for publication May 2013.

0195-9131/13/4512-2353/0
MEDICINE & SCIENCE IN SPORTS & EXERCISE®

DOI: 10.1249/MSS.0b013e31829ba765

However, research in animals reveal that high strain rates (a function of strain magnitude and frequency (5)) rather than high strain magnitudes alone have been identified as the specific mechanism through which dynamic loading positively stimulates the mechanosensory system of bone to result in bone formation (5,20,25,30). Furthermore, Burr et al. (5) concluded that high strain magnitudes are not required to stimulate bone adaptation if the strain rate is sufficiently high. Therefore, the rate at which external force is applied (peak loading rate) provides a more suitable proxy measure for strain rate from which osteogenic thresholds can be developed. For example, a jumping intervention (50 jumps (8.5 cm in height) performed in a single bout 6 $d \cdot wk^{-1}$ for 20 wks) associated with significant increases in BMD in premenopausal women yielded mean peak loading rates of 43 body weights $(BW) \cdot s^{-1}$ per jump landing (1). Given that bone adaptation in response to mechanical stimuli is threshold driven (17), the loading rate presented by Bassey et al. (2) provides an evidence-based value with which to assess whether peak accelerations from hip- and wrist-worn accelerometers can be used as a proxy measure of loading rates that are beneficial to bone.

We recently showed that peak accelerations during walking, running, and jumping activities were positively correlated with mechanical loading in a small sample of young adults (28). Furthermore, the frequency of daily peak accelerations $\geq 3.9g$ and impacts with high acceleration slopes recorded by an accelerometer worn at the hip are reported to be associated with changes in peak BMD in premenopausal women (13,32). The next step is to determine the magnitude of peak accelerations that are associated with the loading rate threshold of 43 $BW \cdot s^{-1}$ across a wide range of typical daily activities. Premenopausal women are of particular interest because of the advantages of maximizing bone before the loss associated with the postmenopausal years (21).

The aims of this study were to a) quantify the relation between peak accelerations assessed at the hip and the wrist and GRF across typical daily activities in premenopausal women, b) determine peak acceleration cut-points that optimize the classification of loading rates above and below a loading rate of 43 $BW \cdot s^{-1}$ for the GENEActiv and ActiGraph GT3X+ accelerometers worn at the hip and at the wrist, and c) determine the sensitivity and specificity of these cut-points for the classification of activities above and below the 43 $BW \cdot s^{-1}$ loading rate threshold in a cross-validation sample.

METHODS

Participants. Forty-seven premenopausal women (age, 39.2 ± 5.6 yr; mass, 65.9 ± 11.0 kg; height, 1.67 ± 0.06 m) age 30–50 yr were recruited from the local area. Each participant filled out a Physical Activity Readiness Questionnaire to confirm her ability to participate in the study. The institutional ethics committee granted approval, and all participants gave written informed consent.

Data collection. After familiarization, each participant performed nine activities in the following order: slow walking, slow walking while carrying a 2-kg shopping bag in the right hand, floor sweeping, brisk walking, slow running, faster running, low jumps, higher jumps, and box drops. For walking, sweeping, and running activities, participants performed 10 shuttles of 12–15 m in length, with GRF data collected for one right step per shuttle. A force plate set flush within the floor (960 Hz; Advanced Mechanical Technology Inc., Watertown, MA) was used to collect GRF data. A metronome was used to ensure the initially self-selected cadence (steps per minute) was consistent between shuttles for activities: slow walk, walking with bags, and floor sweeping, approximately 95 steps per minute; brisk walking, approximately 115 steps per minute; slow running, 130–140 steps per minute; and fast running, 145–160 steps per minute. After familiarization, participants found it easy to keep in time with the step rate and contact the force plate. However, shuttles were discarded if participants failed to make correct contact with the force plate.

Low jumps (approximately 2–5 cm) and higher jumps (>5 cm) were performed continuously (one jump per second) for 20 s on the force plate. Finally, participants dropped from a 20-cm-high box (typical stair height) onto the force plate 10 times. Drops of 20–25 cm have previously been safely used in exercise interventions with postmenopausal women (29). Participants were instructed to land two-footed with bent knees and return to standing to remain stationary on the force plate for 5 s before remounting the box. No restrictions were placed on arm movement throughout the activities. GRF data were analyzed for eight successful steps/jumps for each activity.

Activity monitors. Throughout testing, a waveform GENEActiv (gravity estimator of normal everyday activity) (seismic acceleration sensor, dynamic range ±8*g*, ActivInsights Ltd) accelerometer and an ActiGraph GT3X+ (monolithic differential capacitance sensor, dynamic range ±6*g*, ActiGraph LLC) accelerometer were worn over the right hip (on an elastic waist belt, with the GENEActiv taped onto the ActiGraph) and on the right wrist (accelerometers taped together on a wrist strap). Participants stood still for a four-beat rest at the end of each shuttle to allow accelerometry data for each shuttle to be identified separately. GENEActiv software (version 2.1) and ActiLife5 LITE analysis software were used to initialize the GENEActiv and the ActiGraph, respectively, at a sampling frequency of 100 Hz and to upload the data.

Data analysis. Force plate output variables were peak vertical force expressed as BW (output force/mass (kg) × acceleration due to gravity (9.81 $m \cdot s^{-2}$)) and peak loading rate ($BW \cdot s^{-1}$). For ambulatory activities, steps were viewed individually to check for anomalies before mean GRF output variables were obtained from eight steps using Excel. For jumping activities, a graph containing consecutive vertical force–time histories viewed for each individual indicated that jumps were highly repeatable. Mean GRF output variables for eight trials per jumping activity were therefore

obtained efficiently using a customized MATLAB version R2011b (MathWorks, Natick, MA) code. Using Excel, peak acceleration (g) for each step or jump was extracted from the raw acceleration files for the GENEActiv and the ActiGraph monitors to obtain a mean of eight steps/jumps for each activity. Data for both vertical and resultant accelerations were extracted for the GENEActiv and the ActiGraph worn at the hip, but only data for resultant accelerations were extracted from the GENEActiv and the ActiGraph worn at the wrist. For the monitors worn at the hip, the majority of loading through the body would likely be in line with the vertical vector, but no such assumption can be made for the monitors worn at the wrist.

Statistical analysis. Pearson correlation analysis was used to determine the relation between GRF variables (peak vertical GRF and peak loading rate) and peak acceleration from the accelerometers. Correlations were carried out for all activities combined and for each activity separately.

A calibration sample of 37 participants was randomly selected from the data; the remaining 10 participants formed the cross-validation sample. Receiver operating characteristic (ROC) curves were used to determine peak acceleration cut-points that optimized the sum of sensitivity and specificity for classification of accelerations equating to a peak loading rate >43 $BW \cdot s^{-1}$ (2) for the hip and wrist for both accelerometers. Assuming an area under the curve (AUC) of 0.75, 40 cases above and below the cut-point are needed to provide a power of 0.8 at $\alpha = 0.05$ for comparison with the null hypothesis value (AUC = 0.5) (12). The actual AUC reported for discrimination between activity intensities (classified according to energy expenditure (EE)) has been greater than 0.88 in our previous research (8). In all our ROC analyses, more than 200 cases were above and below the cut-point, meeting this requirement. The sensitivity, specificity, and overall agreement of the resulting accelerometer cut-points were evaluated in a cross-validation with the remaining 10 participants. Performance of the model was also assessed on the basis of leave-one-out cross-validation.

Alpha was set at 0.05. IBM SPSS version 20.0 (IBM, Armonk, NY) was used for all statistical analyses.

RESULTS

Participant characteristics for the whole sample, calibration sample, and cross-validation sample are presented in Table 1. GRF and accelerometer data by activity are presented in Figure 1a–e. Peak vertical GRF was lowest for walking and then increased steadily across slow running, fast running, low jumping, high jumping, and box drops (Fig. 1a). The pattern for peak loading rate differed in that values were lower for low jumping relative to running (Fig. 1b). Peak acceleration at the hip showed a similar pattern to peak loading rate (Fig. 1c–d). Peak acceleration at the wrist again showed a similar pattern to peak loading rate, with the exception of peak wrist acceleration for box drops, which was similar to or lower than that for high jumps (Fig. 1e).

Correlations. The overall pattern of correlations between GRF and accelerations for activities considered separately and activities combined was largely similar irrespective of monitor and, to a lesser extent, irrespective of wear site. Considering all activities together, peak accelerations at the hip and wrist were positively correlated with peak vertical GRF ($r > 0.8$ and $r > 0.7$, respectively, both $P < 0.001$), irrespective of monitor (final column, Table 2). Correlations between peak acceleration and peak loading rate were lower, both at the hip ($r > 0.7$, $P < 0.001$) and at the wrist ($r > 0.57$, $P < 0.001$). The inclusion of multiple data points from each subject for these correlations violates the assumption of independence of observations. Thus, a series of linear mixed effect models was run to assess whether subject had a significant effect on the relations shown in Table 2. There was no significant effect of subject on any of the relations observed.

Considering activities separately, peak accelerations were positively and significantly correlated with peak vertical GRF for low jumps, high jumps, and box drops irrespective of monitor or wear site (r = 0.32–0.68, P < 0.01–0.05) (Table 2). Significant positive correlations were also generally observed for running and fast walking for peak vertical acceleration at the hip and resultant acceleration at the wrist from either accelerometer (r = 0.3/0.4, P < 0.01–0.05). Peak resultant acceleration at the hip did not correlate significantly with peak vertical GRF for running. Peak accelerations were positively and significantly associated with peak loading rate during walking (r = 0.2–0.6, P < 0.01–0.05) and jumping (r = 0.3–0.5, P < 0.01–0.05), but not running. Patterns of association between peak acceleration and peak loading rate were largely independent of monitor or wear site.

Calibration. Resultant and vertical peak accelerations from both the GENEActiv and the ActiGraph GT3X+ accelerometers discriminated between peak loading rates greater and less than 43 $BW \cdot s^{-1}$ with high levels of accuracy, irrespective of wear site (AUC >0.92, $P < 0.001$). The ROC statistics and the cut-points that maximized sensitivity and specificity are presented in Table 3. Sensitivity of the optimal cut-point was high, >92%, with specificity >89% at the hip and >83% at the wrist.

TABLE 1. Participant characteristics for whole sample and for calibration and cross-validation samples separately.

	Whole Sample (N = 47)		Calibration Sample (N = 37)		Cross-validation Sample (N = 10)	
	Mean	SD	Mean	SD	Mean	SD
Age (yr)	39.2	5.6	39.1	5.7	39.4	5.8
Height (m)	1.67	0.06	1.67	0.06	1.67	0.05
Mass (kg)	65.9	11.0	66.6	11.5	63.5	9.3
Body mass index ($kg \cdot m^{-2}$)	23.6	3.8	23.8	4.0	22.7	2.5

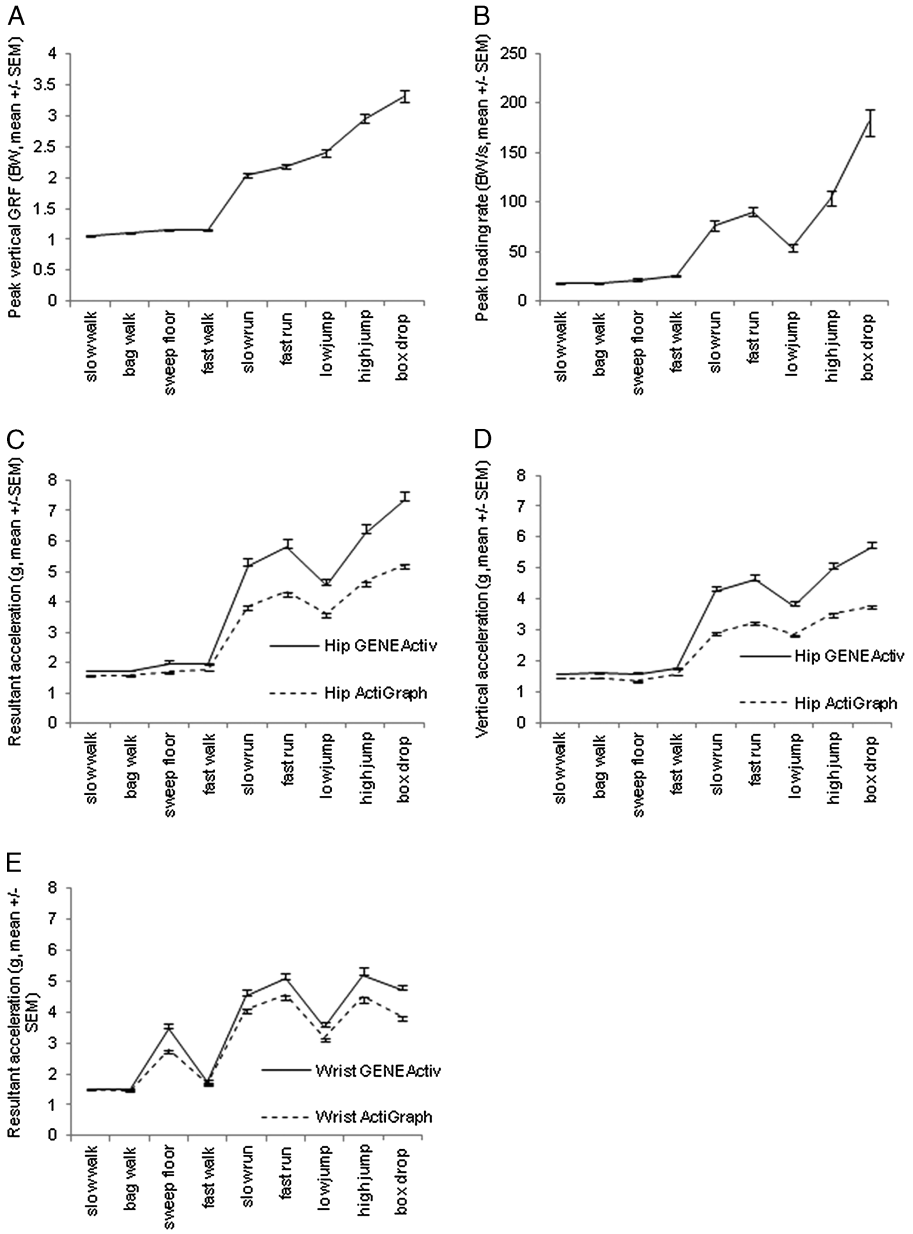

FIGURE 1—Data are presented as mean ± SEM. Peak vertical GRF (BW) (A), peak loading rate ($BW \cdot s^{-1}$) (B), peak resultant acceleration (hip (*g*)) (C), peak vertical acceleration (hip (*g*)) (D), and peak resultant acceleration (wrist (*g*)) by activity (E).

Cross-validation. The cut-points shown in Table 3 were cross-validated in the remaining 10 participants. The sensitivity and specificity for each of the cut-points were >85% (Table 4), with the exception of the specificity for the GENEActiv peak resultant acceleration at the wrist (75%). Overall agreement was >85% for each accelerometer cut-point for the cross-validation sample and for leave-one-out cross-validation.

Figure 2a–f depicts the relations between peak loading rate and peak resultant acceleration at the hip (1a = GENEActiv, 1b = ActiGraph), peak vertical acceleration at the hip (1c = GENEActiv, 1d = ActiGraph), and peak resultant acceleration at the wrist (1e = GENEActiv, 1f = ActiGraph) for the calibration and cross-validation sample. The horizontal lines represent the 43 $BW \cdot s^{-1}$ loading rate threshold, and the vertical lines represent the optimal cut-point for each accelerometer variable. The bottom left and top right quadrants represent correct classification. Data points in the top left quadrant are false negatives, and data points in the bottom right quadrant are false positives.

DISCUSSION

Peak loading rate as a noninvasive estimate of strain rate provides an important measure relevant to bone, but it is not possible to measure peak loading rate during habitual physical activity. The positive relations between peak vertical GRF, peak loading rate, and peak accelerations shown

TABLE 2. Relations between GRF output variables (peak vertical force and peak loading rate) and peak acceleration output (GENEActiv and ActiGraph) by activity and for all activities combined.

			Activity									
Acceleration	Monitor	Location	Slow Walk	Walk with Bag	Sweep Floor	Fast Walk	Slow Run	Fast Run	Low Jump	High Jump	Box Drop	All Activities
Correlations with peak vertical GRF												
Peak resultant *g*	GENEActiv	Hip	0.052	0.352	0.057	0.364*	0.001	0.217	0.648**	0.513**	0.453**	0.835***
	ActiGraph	Hip	0.107	0.299	0.129	0.404**	0.038	0.128	0.660**	0.625**	0.541**	0.878***
	GENEActiv	Wrist	0.178	0.263	0.119	0.283	0.388**	0.372**	0.527**	0.568**	0.498**	0.712***
	ActiGraph	Wrist	0.180	0.214	0.174	0.289*	0.466**	0.422**	0.483**	0.537**	0.393**	0.707***
Peak vertical *g*	GENEActiv	Hip	0.119	0.344*	0.118	0.461**	0.116	0.341*	0.629**	0.663**	0.574**	0.867***
	ActiGraph	Hip	0.224	0.312*	0.082	0.523**	0.391**	0.438**	0.597**	0.681**	0.322*	0.878***
Correlations with vertical peak loading rate												
Peak resultant *g*	GENEActiv	Hip	0.369*	0.446**	0.064	0.523**	0.073	0.111	0.541**	0.501**	0.280	0.714***
	ActiGraph	Hip	0.489**	0.553**	0.054	0.516**	0.021	0.015	0.541**	0.586**	0.394**	0.738***
	GENEActiv	Wrist	0.297*	0.609**	0.066	0.359*	0.086	0.011	0.352*	0.410**	0.368*	0.577***
	ActiGraph	Wrist	0.390**	0.513**	0.007	0.343*	0.149	0.079	0.267	0.400**	0.412**	0.573***
Peak vertical *g*	GENEActiv	Hip	0.278	0.374**	−0.027	0.522**	0.157	0.206	0.532**	0.524**	0.350*	0.732***
	ActiGraph	Hip	0.414**	0.523**	0.097	0.423**	0.202	0.202	0.450**	0.454**	0.251	0.715***

*P < 0.05–0.046.
**P < 0.01.
***P < 0.001.

herein demonstrate that these accelerometers worn at the hip or wrist provide a measure that reflects the short, dynamic bursts of activity known to benefit bone (2,5,13,14,26,32) and can be used during normal habitual activity.

Indeed, the frequency of daily acceleration peaks >3.9*g* recorded by an accelerometer worn at the hip has previously been shown to be associated with changes in peak BMD in premenopausal women (32). The current study has determined peak acceleration cut-points specific to the commercially available GENEActiv and ActiGraph GT3X+ accelerometers that can be used to capture the occurrence of peak accelerations that exceed the loading rate of 43 BW·s^{-1} shown to be beneficial to bone by Bassey et al. (2). Furthermore, the data indicate that this is possible when the accelerometer is worn at the wrist — this is very significant because the wrist is a wear site that is more acceptable to participants and thus results in greater wear time (33).

The optimal cut-points for the hip were greater than those for the wrist, indicating that cut-points need to be specific to the wear site. This supports findings from EE calibration studies (8). At the hip site, there appeared to be no consistent advantage of a triaxial measure of peak acceleration (resultant) over vertical peak acceleration. Some, but not all, studies focusing on relations between EE and acceleration have reported stronger relations when using triaxial accelerometer output (7,9,15,22,27). The lack of an advantage of three axes of measurement in the current study probably reflects the criterion measures of peak vertical GRF, peak loading rate, and the nature of the activities. It is possible that in "real world" or sports activities, resultant accelerations may better reflect the strains on the bone because of movements involving sudden changes of direction and movement in different planes (6). Inclusion of such movements was beyond the scope of the present study partly because of difficulties in a range of participants being able to sufficiently control and safely replicate unfamiliar movements. When an accelerometer is worn at the wrist, it is necessary to have a measure of resultant acceleration because there is no dominant plane of movement. Similar accuracy (>85%) at the two sites for classifying accelerations equivalent to peak loading rates above and below the threshold of 43 BW·s^{-1} supports the use of the wrist-worn accelerometer to assess activity beneficial to bone. This is very encouraging because of the greater acceptability of wrist-wear and thus greater wear time achieved (33). However, the correlations between peak vertical GRF, peak loading rate, and peak accelerations at the wrist were slightly lower than that of the corresponding peak accelerations assessed at the hip. Further research is needed to establish whether relations between bone health and the frequency of peak accelerations above the relevant cut-point during habitual physical activity are similar between wrist- and hip-worn accelerometers.

Correlations between peak vertical GRF, peak loading rate, and peak accelerations were similar between the two brands of

TABLE 3. Calibration: ROC statistics and peak acceleration cut-points for the classification of accelerations associated with loading rates beneficial to bone (>43 BW·s^{-1}).

Monitor	Peak Acceleration	Site	AUC (95% Confidence Interval)	SE	Cut-Point (*g*)	Sensitivity (%)	Specificity (%)
GENEActiv	Resultant	Hip	0.961* (0.938–0.983)	0.012	3.395	94.0	91.5
	Vertical	Hip	0.963* (0.941–0.985)	0.011	3.055	94.6	91.5
	Resultant	Wrist	0.923* (0.892–0.953)	0.016	3.125	92.3	83.0
ActiGraph	Resultant	Hip	0.966* (0.884–0.946)	0.011	2.925	95.2	93.3
	Vertical	Hip	0.962* (0.940–0.984)	0.011	2.170	98.2	89.1
	Resultant	Wrist	0.938* (0.912–0.964)	0.013	2.840	92.9	84.8

N = 37.
*P < 0.001.

TABLE 4. Cross-validation: sensitivity, specificity, and overall agreement associated with the cut-points equivalent to loading rates >43 BW·s^{-1} (Table 3).

			Cross-validation Sample (N = 10)				Leave-One-Out Cross-validation
Monitor	Peak Acceleration	Site	Cut-Point (g)	Sensitivity (%)	Specificity (%)	Overall Agreement (%)	Overall Agreement (%)
GENEActiv	Resultant	Hip	3.395	85.7	85.4	85.6	87.5
	Vertical	Hip	3.055	88.1	85.4	86.7	90.3
	Resultant	Wrist	3.125	97.6	75.0	85.6	85.1
ActiGraph	Resultant	Hip	2.925	85.7	87.5	86.7	90.5
	Vertical	Hip	2.170	97.6	85.4	91.1	89.8
	Resultant	Wrist	2.840	90.5	81.3	85.6	85.3

accelerometer. Classification accuracy during cross-validation was also similar, irrespective of monitor brand or wear site (>85%). This was expected because both monitors are measuring raw accelerations in the same planes and were worn taped together at the respective sites. However, there was a discrepancy evident in the peak acceleration values from the GENEActiv and ActiGraph GT3X+ accelerometers, particularly during higher intensity activities, with the ActiGraph GT3X+ recording lower values than the GENEActiv. The higher dynamic range of the GENEActiv (±8g compared with ±6g for the ActiGraph) may contribute to this discrepancy. This highlights the importance of monitor-specific calibrations, despite a common output measure.

Vainionpää et al. (32) have previously shown that the number of vertical peak accelerations exceeding 3.9g recorded by an accelerometer worn at the hip is associated with changes in BMD in the proximal femur of premenopausal women. These accelerations equate to values of 4.9g in the current study because Vainionpää et al. (32) subtracted 1g from values to account for the acceleration of gravity. This cut-point was not related to mechanical loading or strain but to the outcome of a positive change in BMD. The criterion loading rate used in this study was also related to the outcome of a positive change in BMD (2). However, the acceleration value reported by Vainionpää et al. is considerably higher than the values in the current study. Whether this discrepancy is a real difference in peak accelerations determined by the two different methods or whether it is confounded by differences between the accelerometers themselves is not clear. The vertical acceleration peaks reported by Vainionpää et al. during walking are similar to those derived from the GENEActiv and ActiGraph in our study, but at higher intensities (e.g., box drops), the peak accelerations recorded by the GENEActiv are slightly lower (equivalent to 4.7g ± 1.2g) and those recorded by the ActiGraph much lower (equivalent to 2.8g ± 0.7g) compared with approximately 5.6g presented by Vainionpää et al. (32). This may be because the drops described previously by Vainionpää et al. (31) ranged from the height of a one-step bench (10 cm) up to the height of three benches (30 cm), depending on participant progression through the 12-month exercise intervention (Vainionpää et al. (31)). Therefore, it is not possible to determine the exact magnitude of peak gravity from the study of Vainionpää et al. (32) associated with a drop height of 20 cm as used in this study. These discrepancies in acceleration output again underline the importance of determining monitor-specific cut-points.

Peak accelerations in this study were significantly correlated with peak loading rate across the range of activity intensities and, when looking at individual activities, walking as well as jumping was positively correlated with peak loading rate. This is important because it means that peak acceleration can be used as a proxy measure of strain rate across the intensity range and not just for higher strain activities. Generally, it is reported that high-impact exercise, particularly jumping, is needed for beneficial effects on bone mass (22,23). Consequently, the criterion peak loading rate used in this study to develop the cut-points was based on the association between the mean peak rate of loading experienced by premenopausal women during a jumping intervention that resulted in significant increases in BMD postintervention (2).

However, there have also been several studies that have reported that walking has associations with fracture risk or bone health. For example, postmenopausal women who walked 4 h·wk^{-1} had a 41% reduced hip fracture risk compared with those who walked less than 1 h·wk^{-1}, and those who walked fastest had a 65% lower risk of hip fracture than slower paced walkers (10). Walking has also been found to attenuate bone loss in older men (3), and brisk walking (60 min three times per week at 67%–70% HR_{max}) has been reported to increase BMD in human immunodeficiency virus-infected patients receiving combination antiretroviral therapy (4). Peak loading rate, or strain, is relatively low during walking, but although strain rate has been found to be a clear determinant of cellular activity governing bone adaptation (5), it has also been suggested that the mechanosensory system of bone is sensitive to a variety of mechanical stimulations (11,19). In particular, it has been indicated that low-level stimulations normally "ignored" by bone may become highly anabolic if the temporal pattern of activity increases to higher frequencies (19). Thus, it is possible that relatively low peak accelerations may also be osteogenic if repeated often enough; however, the specific frequency-related mechanisms that may lead to beneficial effects on bone from low-intensity activities are not yet defined. If this is the case, the presence of a correlation between peak loading rate and peak acceleration at the hip and wrist across the whole range of intensities, and not only above a certain threshold, means that these frequency-related mechanisms can be identified. Thus, it is recommended that future studies examining relations between habitual physical activity and bone outcomes not only assess

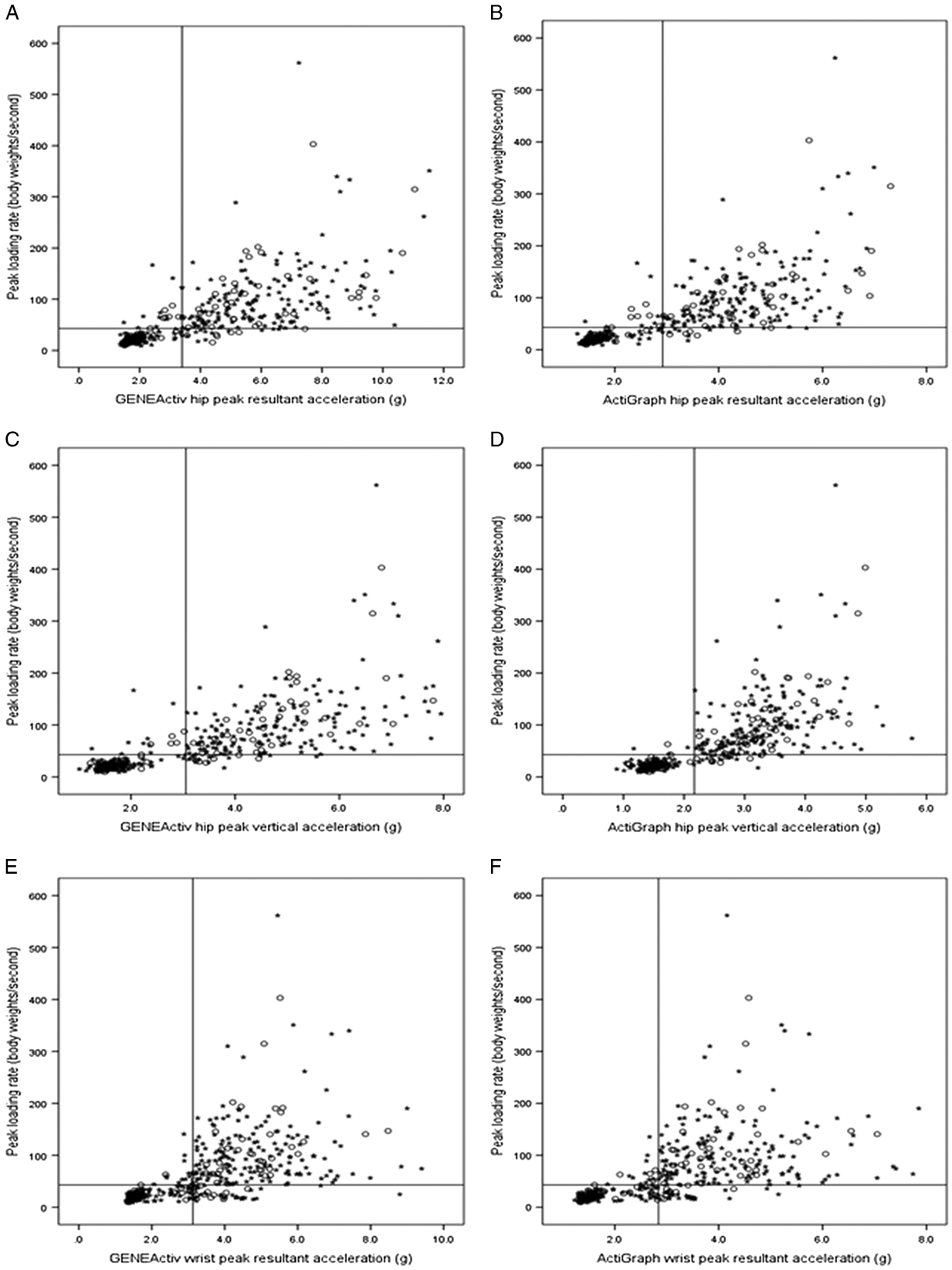

FIGURE 2—Relation between peak loading rate and GENEActiv hip peak resultant acceleration (A), ActiGraph hip peak resultant acceleration (B), GENEActiv hip peak vertical acceleration (C), ActiGraph hip peak vertical acceleration (D), GENEActiv wrist peak resultant acceleration (E), and ActiGraph wrist resultant acceleration (F). *Marked horizontal lines* represent the loading rate of 43 BW·s^{-1} threshold, and *marked vertical lines* represent optimal cut-point for peak acceleration from accelerometer. Calibration group *, Cross-validation group ○.

the frequency and duration of time spent above the peak acceleration cut-points presented but also investigate the effect of differing combinations of magnitude and frequency of peak accelerations on bone health.

A limitation in our previous study was the lack of activities where arm movement was different to leg movement. During normal daily life, many activities take place that involve arm movement that is disassociated from lower body movement. The current study included two such activities: sweeping and walking while carrying bags. Wrist accelerations were elevated in the absence of any increase in loading rate during sweeping. However, they remained below the cut-point associated with the loading rate of 43 $BW \cdot s^{-1}$. It is possible that other confounding activities may lead to more errors in classification when using a wrist-worn accelerometer; e.g., wrist accelerations will be high during arm swinging while standing still and during certain occupational activities, e.g., hammering. Examination of the time spent in different acceleration thresholds as determined by hip and wrist accelerometers worn contemporaneously alongside an independent measure of activity (e.g., observation, doubly labeled water, and use of time instrument) will help determine how much of a confounder this may be during habitual activity measurement.

We based our acceleration cut-points on a peak loading rate of 43 $BW \cdot s^{-1}$. This was the mean peak loading rate experienced by the premenopausal women during the jumping intervention that led to increases in BMD at the trochanter ($P < 0.05$) and at the femoral neck ($P = 0.06$) (2). However, it should be noted that there were no increases in BMD at the lumbar spine in premenopausal women and no effect on BMD at any site in postmenopausal women. Thus, the cut-points identified in this study are based on a single intervention. Furthermore, they are applicable only to premenopausal women and may be only applicable to hip BMD and not to other clinically relevant skeletal regions.

In conclusion, this study has presented cut-points for the GENEActiv and ActiGraph accelerometers worn at the wrist or hip, which can be used to estimate the occurrence of peak loading rates previously shown to be beneficial to bone in premenopausal women. These accelerometry-based activity monitors can be used by researchers on premenopausal women during habitual activity to unobtrusively measure the duration and frequency of periods spent at and above peak acceleration cut-points that reflect loading rates known to benefit bone. Cut-points offer an efficient and simple way to analyze accelerometer data, but we believe it is important not to lose sight of the richness of the data provided. Because peak accelerations are related to peak loading rate during low- as well as high-intensity activities, they may also be used to investigate possible frequency-related mechanisms that may lead to osteogenic benefits from lower intensity activities. Data suggest that a wrist-worn accelerometer performs similarly to a hip-worn accelerometer. Further research is needed to confirm whether this is the case in free-living individuals. The ability to unobtrusively quantify osteogenic components of habitual activity patterns from hip- or wrist-worn monitors may directly inform physical activity interventions aiming to prevent osteoporosis in later life.

The authors would like to thank the Peninsula Research Bank Database, Exeter Clinical Research Facility, for assisting in participant recruitment and Isabel Moore for assistance with analysis of the force plate data. V. H. Stiles and A. V. Rowlands received an Innovative Award from the National Osteoporosis Society, UK (reference number 117/208), to carry out this work. P. J. Griew was employed on the project.

There are no conflicts of interest. The views in this article do not necessarily reflect the views of the National Osteoporosis Society.

The results of this study do not constitute endorsement by the American College of Sports Medicine.

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