International Journal of Applied Sports Sciences
2018, Vol. 30, No. 1, 1-10.
https://doi.org/10.24985/ijass.2018.30.1.1
ISSN 2233-7946 (Online)
ISSN 1598-2939 (Print)

Invited Review Article

Interaction of factors affecting vibration transmission to skeleton during standing: A narrative review

Harshvardhan Singh[1,3*], William R. Reed[1], Donald H. Lein[1], Shannon L. Mathis[2], & Richard D. Davis[3]
[1]*Department of Physical Therapy, University of Alabama at Birmingham, Birmingham, AL, USA*
[2]*Department of Kinesiology, University of Alabama in Huntsville, Huntsville, AL, USA*
[3]*Department of Pediatrics, School of Medicine, University of Alabama at Birmingham, Birmingham, AL, USA*

Abstract

There is a lack of consensus on the effects of vibration therapy on bone outcome measures. Vibration is a mechanical stimulus and can produce mechanical loading on bone. Similar to site-specific effects of mechanical loading on bone, vibration therapy can also produce site-specific effects. Notably, skeletal effects of vibration therapy could depend on the degree of vibration signal that is received by respective skeletal sites. Thus, vibration transmissibility can dictate, in part, effects of vibration therapy on bone outcome measures. Factors at various levels such as the type of vibration, type of population receiving vibration, and their interaction could affect vibration transmission. In addition, vibration amplitude, vibration frequency, joint position, body posture, resonance frequency of skeletal sites, tissue composition of human body including bone geometry can affect vibration transmission across the human body. The main aim of this review is to summarize the published evidence of various factors that affect vibration transmission which will help to inform future evidence based vibration therapy protocols for skeletal rehabilitation in various populations.

Key words: Vibration, Resonance, Bone, Posture, Aging, Accelerometer

Introduction

There is evidence that vibration has anabolic potential for bone in various populations (Gilsanz et al., 2006; Reyes, Hernández, Holmgren, Sanhueza, & Escobar, 2011; Rubin et al., 2004; Ward et al., 2004; Wren et al., 2011). With the knowledge that lower levels of physical activity are seen in children with movement disorders (Carlon, Taylor, Dodd, & Shields, 2013; Mitchell, Ziviani, & Boyd, 2015) which can lead to underdeveloped bone, and that lack of physical activity is associated with deficits in bone density across childhood to adulthood (Koedijk et al., 2017), nonpharmacologic treatments that could attenuate adverse effects of reduced physical activity are clinically important. Recently, high-frequency, low-intensity vibration (HLV), which is any vibration with a frequency in the range of 30-100 Hz and an intensity of less than 1 g, (where g is the earth's gravitational field), has been shown to have anabolic potential on bone in various clinical subpopulations including clinical populations such as children with cerebral palsy (CP) (Ward et al., 2004; Wren et al., 2011).

In a seminal research study, trabecular bone density, trabecular bone volume and trabecular bone number were

Submitted : 7 June 2018
Revised : 14 June 2018
Accepted : 21 June 2018
Correspondence : hsingh@uab.edu

found to be ~30-45% greater in HLV-treated sheep compared to controls (Rubin, Turner, Bain, Mallinckrodt, & McLeod, 2001). However, clinical trials of vibration effects on skeletal parameters have been mixed. Furthermore, a clinical trial in children with CP showed beneficial effects of HLV on trabecular volumetric bone mineral density (BMD) at the tibia in response to a daily standing on an HLV platform (Ward et al., 2004). Positive effects of HLV on bone has been reported in postmenopausal women (Rubin et al., 2004), young women (Gilsanz et al., 2006), and children with various motor disabilities (Reyes et al., 2011) suggesting HLV is anabolic to bone in various populations. However, there is some evidence that HLV may not be an effective intervention for increasing bone related outcome measures (Leung et al., 2014; Slatkovska et al., 2011) or HLV may produce inconsistent effects across different skeletal sites (Lam et al., 2013). A randomized clinical trial found no improvements in trabecular volumetric BMD or femoral neck, lumbar spine, or the total hip BMD in osteopenic postmenopausal women although adherence to HLV was poor in this study (Slatkovska et al., 2011).

Notably, the exact mechanism by which HLV acts on bone is unclear. Current evidence suggests that mechanical loading at cellular level (Schlueter et al., 2006), stochastic resonance (Tanaka, Alam, & Turner, 2003), decreased osteoclast activity (Xie et al., 2006),and increased bone formation rate (Oxlund, Ørtoft, Andreassen, & Oxlund, 2003) could dictate, in part, effects of HLV on bone. Although mechanical loading has site-specific influence on bone with greatest effects at the site that experiences the most load (Haapasalo et al., 1998; Kannus, 1995), it is not clear if the anabolic effects of HLV are site-specific. Based on Frost's mechanostat model (Frost, 2003), it can be postulated that adaptation of bone to HLV may be related to the degree of vibration transmission to a specific site. Therefore, it is important to understand the factors which could affect vibration transmission to skeleton, specifically during standing which is the most common posture in which the vibration intervention is performed.

Factors affecting vibration transmission in standing can be grouped in three categories: 1) source vibration acceleration, 2) the human body, and 3) the interaction between human body and vibration acceleration source (Harazin & Grzesik, 1998). The main aim of this review is to provide a mechanistic framework to understand various factors that interplay to affect vibration transmission during standing.

Interaction of posture, joint position, vibration amplitude, and vibration frequency affects vibration transmission

Body posture is one of the critical factors affecting vibration transmission during standing (Harazin & Grzesik, 1998) and different joint positions lead to different postures. In addition, different postures can impact the 1) position of the bony site receiving vibration acceleration, 2) level of muscular tension or intramuscular pressure in different muscle groups, and 3) contact area of feet with the accelerating source. Additionally, different postures can influence local site-specific resonance, which is the tendency of a system to vibrate at a greater amplitude at some specific frequencies or a range of frequencies, at a higher frequency leading to a variable response in vibration transmission in unique frequency bands (Griffin & Matsumoto, 1998; Harazin & Grzesik, 1998; Zhao, Dodge, Nemani, & Yokota, 2014).

Although, vibration transmission in seated position has been established (Paddan & Griffin, 1998) not much is known about vibration transmission in standing position, especially with respect to HLV transmission, at different skeletal sites. Most of the earlier work investigated vibration transmission to the head (Paddan & Griffin, 1998) in the standing position. Garg et al. (Garg & Ross, 1976) measured vibration acceleration transmissibility to the head in the frequency range of 1 - 50 Hz. and 0.003 - 0.02 mm in amplitude. In this study, 12 healthy adult subjects (average age 23.42 years) were tested and individual plots were obtained for each subject. Resonance peaks around 2, 6, 20, and 40 Hz were noted in all 12 plots. Garg and Ross (Garg & Ross, 1976) inferred that the effective mass of the body on specific skeletal sites could play a critical role in dictating resonance frequency of respective skeletal sites.

An interaction between posture and vibration magnitude was reported in a study by Matsumoto et al (Griffin & Matsumoto, 1998). Their study investigated vibration transmission at various sites such as T1, T8, L4 spines, iliac crest, and knee (i.e., patella) in 12 healthy adult males (age range: 24 - 35 years) in response to vibration acceleration over a frequency range of 0.5 - 30 Hz and an amplitude of 0.125 - 2 m/s^2 root mean square (0.017 - 0.2 g). Accelerometers were used to measure vibration transmission in various postures - normal standing, legs bent, and standing on one leg. In this study, the researchers reported an amplification of vibration acceleration at the knee at frequencies greater than 10 Hz in the normal standing posture and at greater than 15 Hz in legs bent posture, whereas, attenuation was noted with one leg standing posture at frequencies less than 20 Hz. Additionally, vertical transmission to the knee (i.e., patella) increased with increasing frequency and increasing magnitude ($p < 0.05$). The findings from this study implied that the human body responds dynamically to the vibration acceleration stimulus and that vibration transmission can be affected by interaction between joint position, body posture, and vibration parameters such as amplitude and frequency.

Harazin et al. (Harazin & Grzesik, 1998) examined the dissemination of vertical vibration transmission (4m/s^2 r.m.s or 0.4g and 4 - 300 Hz) to different parts of the body in various postures in college-aged population (n = 10; mean age = 23.1 years). They measured vibration transmission by accelerometer at various skeletal sites such as medial malleolus (ankle), lateral epicondyle (knee), iliac crest (hip), metatarsus (foot) and acromion process (shoulder). The transmissibility curve at the medial malleolus showed resonance at three bands: 4 - 8, 12.5, and 25 - 63 Hz. It was shown that around 31.5 Hz, transmissibility at the ankle was more than the unity for all postures except standing on toes posture. On the other hand, resonance at two bands: 4 - 8 and 12.5 - 25 Hz were noted at the lateral epicondyle. Also, the lateral epicondyle displayed transmissibility near unity at 31.5 Hz in step standing posture. At frequencies above 31.5 Hz, the lateral epicondyle demonstrated an attenuated vibration transmission in all postures. It should be noted that they reported inter-subject variations in transmissibility to be as high as 6:1 for the vibration transmission from the vibrating source to the ankle and the hip. Also, a set of ten transfer functions corresponding to 10 postures was obtained by Harazin et al (Harazin & Grzesik, 1998) and was used for correction of measure of transmission. Importantly, it has been shown by Kim et al. (Kim, Voloshin, Johnson, & Simkin, 1993) that without transfer functions, skin mounted accelerometer can overvalue the actual peak vibration acceleration by 10 – 15%.

Kiiski et al. (Kiiski, Heinonen, Järvinen, Kannus, & Sievänen, 2008) examined vibration transmissibility in healthy male participants (n = 4; age = 24 - 47 years) over a wide range of vibration amplitude (0.04 g - 19 g) and frequencies (10 - 90 Hz) using skin mounted accelerometers (20 grams) at the ankle (medial malleolus) and knee (tibial tuberosity). Every participant in their study reported discomfort between 20 - 25 Hz frequency at amplitude of 0.5 mm or greater (i.e., 0.8 and 7.5 g). Sizeable amplification of peak vibration acceleration between 10 - 40 Hz and 10 - 25 Hz was observed at the ankle and knee, respectively. Also, waveforms emitted by the vibrating platform appeared distorted and lost their sinusoidal character with higher magnitude vibration acceleration. Thus, not only the vibration amplitude but also the weight of accelerometers which would affect the accelerometers inertia to move and propagate vibration, comfort of patient, and the degree of tightness of attachment of the accelerometer to the skin at respective bony sites can also affect vibration transmission.

To our knowledge, only two studies (Bressel, Smith, & Branscomb, 2010; Singh et al., 2017) have examined vibration transmission in children. Bressel et al. (Bressel et al., 2010) investigated the nature of vibration transmission in typically developing children (n = 11) versus adults (n = 10). Motion analysis system was used to measure vibration transmission when subjects stood on the vibrating platform (28/33/42 Hz). Although, transmissibility at the head was not different between children and adults ($p = 0.92$), transmissibility in children was 42% and 62% greater than adults for the ankle (lateral malleolus) and the hip (anterior superior iliac spine) ($p = 0.03$), respectively. It was noted that vibration accelerations at all the sites increased with frequency and magnitude. At the ankle, transmissibility was found to be two times higher than the vibration acceleration at the platform. However, at the knee (tibial tuberosity), vibration acceleration was attenuated in children at all the frequencies. A novel finding of this study was amplified values of vibration acceleration at 33 Hz setting which was higher than 28 Hz. Researchers credited this amplification of vibration transmission at the distal tibia to the resonance frequency of the distal tibia which is ~30 Hz. Similar to the findings of this study, Singh et al. (Singh et al., 2017) reported an amplified signal transmission at the ankle (medial malleolus) than at the platform in typically developing children and children with CP. However, HLV signal at the knee (lateral condyle) attenuated in children with cerebral palsy but remained amplified in typically developing children. One of the reasons why the HLV signal showed amplification (Singh et al., 2017) versus attenuation (Bressel et al., 2010) at the knee could be due to the site selection. Singh et al. (Singh et al., 2017) measured vibration transmission at the knee at the lateral femoral condyle while Bressel et al. (Bressel et al., 2010) used lateral tibial tuberosity to measure vibration transmission at the knee. Different tissue composition of the bone along its length and the fact that resonance frequency is altered along the length of bone (Zhao et al., 2014) may have contributed to these results. Singh et al. (Singh et al., 2017) also reported an inverse relationship between the level of spasticity of lower leg muscles and vibration transmission at the knee. Spasticity may affect the standing posture which, in turn, may affect vibration transmission.

It should be noted that human body responds dynamically to any external stimulus. Also, the physiological response to any external stimulus usually is non-linear. Mansfield et al. (Mansfield & Griffin, 2000) showed that human body exhibits a non-linear biodynamic response to vertical vibration. In their study, 12 subjects were exposed to vibration stimulus [0.25 (0.03g) - 2.5 (0.26g) ms^{-2}] in the frequency range of 0.2 - 20 Hz in a seated position. Vibration transmission was measured at the upper and lower abdominal wall, L3 spine, posterior superior iliac spine and iliac crest. Resonance frequency for the apparent mass decreased with increased vibration magnitude in all the subjects and apparent mass non-linearities were displayed in the frequency range of 3 - 16 Hz. Furthermore, frequency of the first resonance decreased with increase in vibration magnitude and the shape of graphs depicting vertical transmission from seat to the spine and pelvis showed a general non-linear response. A non-linear response could be related to the unique geometrical shape of the human skeletal system where bony segments are non-linear in shape which may dictate, in part, a non-linear response to vibration transmission.

An interesting study (Prinz, 1997) showed the importance of joint position in vibration transmission in a temporomandibular joint. In this study (Prinz, 1997), the relative orientation of two contributing bones that constitute a joint was found to affect vibration transmission. Furthermore, ligaments which bind one bone to another bone also changed their length with different joint positions such as open (when the ligaments became tight) versus closed jaw (when the ligaments became slack) which might have contributed to different patterns of vibration transmission with different positions (high vibration transmission with the tight ligaments and lower vibration transmission with slack ligaments). Since the ankle and the knee joint can be considered hinge joints (Scott & Winter, 1991) similar to the temporomandibular join, it can be postulated that joint position can play an important role in HLV transmission in these populations.

Technical factors affecting vibration transmission

There are two main techniques of measuring vibration transmission: invasive (i.e., bone mounted accelerometer), and non-invasive (i.e. skin-mounted accelerometer). Bone mounted accelerometers are considered the ideal technique to assess vibration transmission. Previous study has used accelerometers attached to a pin which was glued to the bone of interest (Rubin et al., 2003) however, skin mounted accelerometers have also been shown to yield reasonable estimates of vibration transmission (Harazin & Grzesik, 1998; Kiiski et al., 2008; Kim et al., 1993). A loss of vibration signal at high frequency component (> 110 Hz) with skin mounted accelerometers compared to bone mounted accelerometers has been reported before (Kim et al., 1993). In addition, between 15 - 30 Hz frequency spectrum, without the transfer-function correction, skin mounted accelerometer can overestimate the bone mounted accelerometer values by ~12% (Kim et al., 1993). However, it should be noted that the nature of vibration assessed in the study by Kim et al. (Kim et al., 1993) was not sinusoidal; rather it was an impact force. Previous studies (Harazin & Grzesik, 1998; Rubin et al., 2003; Singh et al., 2017) have utilized skin mounted accelerometers of various masses, which in turn could alter propagation of vibration transmission. While a previous study used the transfer function correction (Harazin & Grzesik, 1998) to estimate precise vibration transmission via skin-mounted accelerometers; another study (Kiiski et al., 2008) failed to do so. Interestingly, the findings of these studies were similar with respect to transmissibility at various skeletal sites such as the tibial tuberosity and the medial/lateral malleolus, suggesting a wider application of skin mounted accelerometers in research settings is acceptable. In support of this observation, Singh et al. (Singh et al., 2017) reported same pattern of vibration transmission with and without the transfer function correction.

Tissue composition, shape of bone, and vibration transmission

It is plausible that tissue composition could play a critical role in the vibration transmission. It is known that resonance can lead to amplification of the signals when the signal frequency is close to the natural frequency of the tissue. On the other hand, tissues such as fat, muscle and bone can act as low pass filters to dampen the vibration acceleration signal (Wakeling & Nigg, 2001). Notably, tissue composition could be postulated to affect vibration transmission in a site-specific pattern during standing posture. This is not entirely unexpected because factors that could affect vibration transmission such as muscle, adiposity, and other soft tissues such as ligament are not present in appreciable amount between ankle, foot and the source of vibration. It is only at the sites above the medial/lateral malleolus that the effect of soft tissues on vibration transmission can be postulated to alter the vibration transmission to any measurable degree..

Previous studies (Dodge et al., 2012; Wakeling & Nigg, 2001) have emphasized the role of muscles in damping of the vibration acceleration signals. Wakeling et al. (Wakeling & Nigg, 2001) determined the role of muscle length, muscle contraction velocity, and muscle force production in damping the vibration signals. Accelerometers (15.9 grams) were used to measure vibration acceleration at various muscles, such as gastrocnemius, tibialis anterior, and vastus lateralis in 14 young adults (male = 7, female = 7). Isometric muscle torques, joint angle and joint angular velocity were recorded by a Biodex apparatus. Vibration stimulus was provided to muscle during each contraction by a strike with a wooden mallet. Increased damping was observed with an increase in torque for any angle and increasing joint power for all directions (x, y, and z). Results from this study suggested that the damping coefficient of the oscillations increased with increased muscular force production. An activated muscle generates tension which in turn depends on the number of recruited motor units eventually resulting in the generation of intermuscular pressure. Additionally, fascia can also contribute toward the enhancement of intermuscular pressure. An individual difference in muscle geometry, muscle-tendon compliance andthe pattern of recruitment between motor units can affect intramuscular pressure which in turn can affect vibration propagation.

Reduced number of motor units (Marciniak, Li, & Zhou, 2015) and reduced motor unit recruitment (Xu, Mai, He, Yan, & Chen, 2015) in the affected versus the unaffected side in children with CP can adversely affect the capacity of muscle to dampen the vibration stimulus. It can be postulated that an interplay of various factors such as muscle mass, muscle quality, neuromuscular feedback, adiposity, and blood pressure could be the driving force in determining intermuscular pressure, the level of which can affect damping of vibration accelerations. Wakeling et al. (Wakeling & Nigg, 2001) also showed that damping coefficient depends on absolute mass of soft tissue such as muscle. An increased muscle mass was found to decrease the damping coefficient. Additionally, an independent effect of joint angle was reported on damping of vibration acceleration signals. An increased damping coefficient was reported with an increased muscle contraction velocity. Currently, the response of muscle tuning to HLV transmission remains unknown. It can be postulated that factors such as adipose tissue infiltration of muscle, muscle length, and musculo-tendinous compliance could affect muscle contraction velocity and thus, could interact with other factors affecting vibration transmission. However, this postulation remains to be investigated.

Inaddition to muscle, the collagen content of bone can be a critical factor that can affect vibration transmission. In one of the seminal studies, the collagen content of the femur bone was estimated across different species, such as ox, guinea pig, and human (Jee, 1983). The study utilized samples of freshly excised bones from oxen and guinea pigs, while in humans, bone samples were taken from the mid-femur from children and adults across a wide-spectrum of age after autopsies (9 months to 90 years). The collagen content was found to be ~90% - 95% of the total organic matrix across different species. Mineralization was found to increase with age in humans. Since collagen provides the ductility and ability to absorb external energy/shock, its role in HLV transmission could be critical.

In a study (Dequeker & Merlevede, 1971) done on human trabecular bone authors provided evidence of increasing collagen content with increasing age, increasing porosity, and decreasing bone mass. Authors studied the collagen content in trabecular bone from the iliac crest that was obtained after autopsy in the age range of 23-83 years. The organic matrix represented 28% by weight of the fat-free dry bone, and collagen 23%. Thus, the collagen content constituted 80% of the organic matrix. A tendency of increased collagen content was noted in the more porotic samples. A significant amount of collagen becomes insoluble with increasing age and increasing porosity was a novel finding of this study. A change in the nature of collagen cross-links can be hypothesized to affect vibration transmission. These findings can be related to a work by Currey et al. (Currey, 1979) where the authors found a decrease in impact absorption energy by more than three-fold between the ages of three and ninety. Furthermore, mineralization was higher in older bone. Interestingly, higher porosity in young bones did not materially affect its energy absorption but the higher porosity of older bones produced a deleterious effect on impact energy absorption.

Shape and mineral distribution of bone can also be a critical factor affecting vibration transmission. A recent study (Campoli et al., 2014) based on finite element modeling reported that shape and mineral distribution in human femur can induce large variations in the calculated natural frequency/resonance frequency. Twenty-two different modes of femoral shape were created to describe 95% of the variance in the femoral appearance. In this study (Campoli et al., 2014), changing the femoral shape from the mean shape by up to three standard deviations led to 7-16% variation in the natural frequencies of the vibration. Overall, natural frequencies varied between 25 –38% from their mean values for all the modes that were tested. This implies that vibration of a bone to induce its maximum apposition by choosing a specific frequency as a resonance frequency as suggested by Zhao et al. (Zhao et al., 2014) is not recommended.

Patient-specificity with respect to body composition can likely play a critical role in imparting resonance frequency to the femur. A recent study by Dodge et al. (Dodge et al., 2012) examined damping capacity of bone, joint tissue, muscle, and skin in the lower hind limb of mice (n = 25) in response to a 5 N peak–to–peak force applied at 0.5 Hz increments from 1-20 Hz. Two damping parameters (i.e., phase shift angle and dissipative energy) were determined in 5 different samples: 1) intact lower hind limb, 2) limb without the skin, 3) limb without the skin and foot, 4) the tibia and the fibula alone without an intact joint at either end or 5) muscle removed. A significant difference was noted in phase shift angle and dissipative energy in sample 4 and 5 than the others ($p < 0.001$). Data from their study showed that muscle, bone, and knee joint were the largest contributors to the energy loss in response to axial loading while skin had no effect on damping at any the frequencies. Interestingly, normal bone curvature enhanced the damping capacity of bone by 40%. The findings from this study suggested that the bone curvature, the bone's damping capacity, and the muscle surrounding the bone are the biggest contributors to the external mechanical damping. However, the researchers did not investigate the effect of leg adiposity on mechanical damping.

Recently, Zhao et al. (Zhao et al., 2014) showed that Young's modulus can affect the resonance frequency of a skeletal site suggesting factors such as age, gender, and bone structure could modulate a skeletal site's response to loading at specific frequencies. The study by Zhao et al. (Zhao et al., 2014) also suggested that an enhanced quality of trabecular bone microarchitecture will be related with better vibration transmission and vice versa suggesting an important role of bone quality in the vibration transmission. Interestingly, bones in children have low Young's modulus and yet absorb more energy before sustaining facture. To our knowledge, no one has looked at the role of adiposity, muscle volume, and trabecular bone microarchitecture in vibration transmission in children with CP. also, with the knowledge that children with CP show an infiltration of adipose tissue in the muscles of their lower extremity, and have underdeveloped trabecular bonemicroarchitecture, it

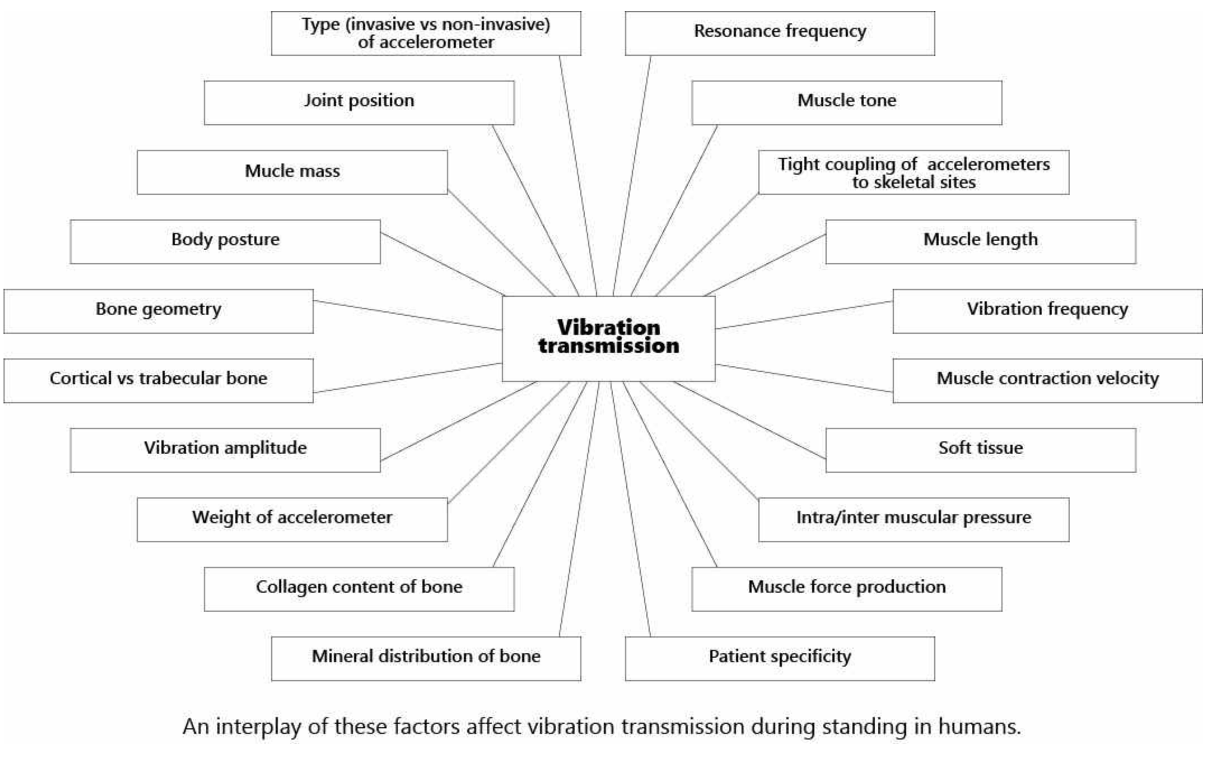

Figure 1: *Factors affecting vibration transmission*

becomes pertinent to know how tissue composition can affect HLV transmission, and if that affects the adaptation of bone to HLV transmission.

Conclusion

Overall, previous evidence indicates that vibration transmission across skeletal sites in humans depend on various factors such as age, geometry of bone, skeletal site, soft tissue surrounding bone, joint position, body posture, vibration amplitude, vibration frequency, and type of accelerometer used to assess vibration transmission as shown in Figure 1. These factors may be related to inconsistent effects being reported in previous literature on effects of vibration on skeletal parameters. Studies focusing on effects of vibration transmission on skeletal parameters should report the degree of transmission at respective skeletal sites and should be cautious while inferring the quantity of vibration transmitted at respective skeletal sites.

References

Bressel, E., Smith, G., & Branscomb, J. (2010). Transmission of whole body vibration in children while standing. *Clinical Biomechanics*, **25**(2), 181–186. doi: 10.1016/j.clinbiomech.2009.10.016.

Campoli, G., Baka, N., Kaptein, B. L., Valstar, E. R., Zachow, S., Weinans, H., & Zadpoor, A. A. (2014). Relationship between the shape and density distribution of the femur and its natural frequencies of vibration. *Journal of Biomechanics*, **47**(13), 3334–3343. doi: 10.1016/j.jbiomech.2014.08.008.

Carlon, S. L., Taylor, N. F., Dodd, K. J., & Shields, N. (2013). Differences in habitual physical activity levels of young people with cerebral palsy and their typically developing peers: A systematic review. *Disability and Rehabilitation*, **35**(8), 647–655. doi: 10.3109/09638288.2012.715721

Currey, J. D. (1979). Changes in the impact energy absorption of bone with age. *J.Biomech.*, **12**(6), 459–469.

Dodge, T., Wanis, M., Ayoub, R., Zhao, L., Watts, N. B., Bhattacharya, A., ... Yokota, H. (2012). Mechanical loading, damping, and load-driven bone formation in mouse tibiae. *Bone*, **51**(4), 810–818. doi: 10.1016/j.bone.2012.07.021.

Frost, H. M. (2003). Bone's mechanostat: A 2003 update. The Anatomical Record. Part A, *Discoveries in Molecular, Cellular, and Evolutionary Biology*, **275**(2), 1081–1101. doi: 10.1002/ar.a.10119.

Garg, D. P., & Ross, M. A. (1976). Vertical Mode Human Body Vibration Transmissibility. *IEEE Transactions on Systems, Man, and Cybernetics*, **SMC-6**(2), 102–112. doi: 10.1109/TSMC.1976.5409180.

Gilsanz, V., Wren, T. A. L., Sanchez, M., Dorey, F., Judex, S., & Rubin, C. (2006). Low-level, high-frequency mechanical signals enhance musculoskeletal development of young women with low BMD. *Journal of Bone and Mineral Research*, **21**(9), 1464–1474. doi: 10.1359/jbmr.060612.

Griffin, M. J., & Matsumoto, Y. (1998). Dynamic Response of the Standing Human Body Exposed To Vertical Vibration: Influence of Posture and Vibration Magnitude. *Journal of Sound and Vibration*, **212**(1), 85–107. doi: 10.1006/jsvi.1997.1376.

Haapasalo, H., Kannus, P., Sievänen, H., Pasanen, M., Uusi-Rasi, K., Heinonen, A., ... Vuori, I. (1998). Effect of long-term unilateral activity on bone mineral density of female junior tennis players. *Journal of Bone and Mineral Research*, **13**(2), 310–319. doi: 10.1359/jbmr.1998.13.2.310.

Harazin, B., & Grzesik, J. (1998). The transmission of vertical whole-body vibration to the body segments of standing subjects. *Journal of Sound and Vibration*, **215**(4), 775–787. doi: 10.1006/jsvi.1998.1675.

Jee, W. S. S. (1983). The skeletal tissues. *Histology*, **50**(1948), 200–255.

Kannus, P. (1995). Effect of Starting Age of Physical Activity on Bone Mass in the Dominant Arm of Tennis and Squash Players. *Annals of Internal Medicine*, **123**(1), 27. doi: 10.7326/0003-4819-123-1-199507010-00003.

Kiiski, J., Heinonen, A., Järvinen, T. L., Kannus, P., & Sievänen, H. (2008). Transmission of vertical whole body vibration to the human body. *Journal of Bone and Mineral Research*, **23**(8), 1318–1325. doi: 10.1359/jbmr.080315.

Kim, W., Voloshin, A. S., Johnson, S. H., & Simkin, A. (1993). Measurement of the impulsive bone motion by skin-mounted accelerometers. *Journal of Biomechanical Engineering*, **115**(1), 47–52. doi: 10.1115/1.2895470.

Koedijk, J. B., van Rijswijk, J., Oranje, W. A., van den Bergh, J. P., Bours, S. P., Savelberg, H. H., & Schaper, N. C. (2017). Sedentary behaviour and bone health in children, adolescents and young adults: A systematic review. *Osteoporosis International*, **28**(9), 2507–2519. doi: 10.1007/s00198-017-4076-2.

Lam, T. P., Ng, B. K. W., Cheung, L. W. H., Lee, K. M., Qin, L., & Cheng, J. C. Y. (2013). Effect of whole body vibration (WBV) therapy on bone density and bone quality in osteopenic girls with adolescent idiopathic scoliosis: A randomized, controlled trial. *Osteoporosis International*, **24**(5), 1623–1636. doi: 10.1007/s00198-012-2144-1.

Leung, K. S., Li, C. Y., Tse, Y. K., Choy, T. K., Leung, P. C., Hung, V. W. Y., ... Cheung, W. H. (2014). Effects of 18-month low-magnitude high-frequency vibration on fall rate and fracture risks in 710 community elderly - A cluster-randomized controlled trial. *Osteoporosis International*, **25**(6), 1785–1795. doi: 10.1007/s00198-014-2693-6.

Mansfield, N. J., & Griffin, M. J. (2000). Non-linearities in apparent mass and transmissibility during exposure to whole-body vertical vibration. *Journal of Biomechanics*, **33**(8), 933–941. doi: 10.1016/S0021-9290(00)00052-X.

Marciniak, C., Li, X., & Zhou, P. (2015). An examination of motor unit number index in adults with cerebral palsy. Journal of Electromyography and Kinesiology, **25**(3), 444–450. doi: 10.1016/j.jelekin.2015.02.007.

Mitchell, L. E., Ziviani, J., & Boyd, R. N. (2015). Physical activity for children and adolescents with cerebral palsy: Are they doing enough? *Physical Therapy*, **95**(2), 202–211.

Oxlund, B. S., Ørtoft, G., Andreassen, T. T., & Oxlund, H. (2003). Low-intensity, high-frequency vibration appears to prevent the decrease in strength of the femur and tibia associated with ovariectomy of adult rats. *Bone*, **32**(1), 69–77. doi: 10.1016/S8756-3282(02)00916-X.

Paddan, G. S., Griffin, M. J. (1998). A review of the transmission of translational seat vibration to the head. *Journal of Sound and Vibration*, **215**(4), 863–882. doi: 10.1006/jsvi.1998.1592.

Prinz, J. F. (1997). Vibration transmission across the temporomandibular joint. *Journal of Oral Rehabilitation*, **24**(3), 224–229.

Reyes, M. L., Hernández, M., Holmgren, L. J., Sanhueza, E., & Escobar, R. G. (2011). High-frequency, low-intensity vibrations increase bone mass and muscle strength in upper limbs, improving autonomy in disabled children. *Journal of Bone and Mineral Research*, **26**(8), 1759–1766. doi: 10.1002/jbmr.402.

Rubin, C., Turner, A. S., Bain, S., Mallinckrodt, C., & McLeod, K. (2001). Low mechanical signals strengthen long bones. *Nature*, **412**(6847), 603-604.

Rubin, C., Pope, M., Fritton, J. C., Magnusson, M., Hansson, T., & McLeod, K. (2003). Transmissibility of 15-Hertz to 35-Hertz Vibrations to the Human Hip and Lumbar Spine: Determining the Physiologic Feasibility of Delivering Low-Level Anabolic Mechanical Stimuli to Skeletal Regions at Greatest Risk of Fracture because of Osteoporosis. *Spine*, **28**(23), 2621–2627. doi: 10.1097/01.BRS.0000102682.61791.C9.

Rubin, C. T., Recker, R., Cullen, D., Ryaby, J., McCabe, J., & McLeod, K. (2004). Prevention of postmenopausal bone loss by a low-magnitude, high-frequency mechanical stimuli: A clinical trial assessing compliance, efficacy, and safety. *Journal of Bone and Mineral Research*, **19**(3), 343–351. doi: 10.1359/JBMR.0301251.

Schlueter, P. J., Royer, T., Farah, M. H., Laser, B., Chan, S. J., Steiner, D. F., & Duan, C. (2006). Gene duplication and functional divergence of the zebrafish insulin-like growth factor 1 receptors. *The FASEB Journal*, **20**(8), 1230–1232. doi: 10.1096/fj.05.

Scott, S. H., & Winter, D. A. (1991). Talocrural and Talocalcaneal Joint Kinematics and Kinetics During the Stance Phase of Walking. *Journal of Biomechanics*, **24**(8), 743–752.

Singh, H., Whitney, D. W., Knight, C., Miller, F., Manal, K., & Modlesky, C. M. (2016) Site-specific transmission of a floor-based high frequency, low magnitude vibration stimuus in children with spastic cerebral palsy. *Arch Phy Med Rehabil*. **97**(2), 218-223.

Slatkovska, L., Alibhai, S. M. H., Beyene, J., Hu, H., Demaras, A., & Cheung, A. M. (2011). Effect of 12 Months of Whole-Body Vibration Therapy on Bone Density and Structure in Postmenopausal Women. *Annals of Internal Medicine*, **155**(10), 668–679. doi: 10.1059/0003-4819-155-10-201111150-00005

Tanaka, S. M., Alam, I. M., Turner, C. H. (2003) Stochastic resonance in osteogenic response to mechanical loading. *FASEB J*. **17**(2), 313-314.

Wakeling, J. M., & Nigg, B. M. (2001). Modification of soft tissue vibrations in the leg by muscular activity. *J Appl Physiol*, **90**(2), 412–420. doi: 10.1152/jappl.2001.90.2.412.

Ward, K., Alsop, C., Caulton, J., Rubin, C., Adams, J., & Mughal, Z. (2004). Low magnitude mechanical loading is osteogenic in children with disabling conditions. *Journal of Bone and Mineral Research*, **19**(3), 360–369. doi: 10.1359/JBMR.040129.

Wren, T. A. L., Ph, D., Lee, D. C., Hara, R., Susan, A., Kay, R. M., … Gilsanz, V. (2011). *NIH Public Access*, **30**(7), 732–738. doi: 10.1097/BPO.0b013e3181efbabc.

Xie, L., Jacobson, J. M., Choi, E. S., Busa, B., Donahue, L. R., Miller, L. M., … Judex, S. (2006). Low-level mechanical vibrations can influence bone resorption and bone formation in the growing skeleton. *Bone*, **39**(5), 1059–1066. doi: 10.1016/j.bone.2006.05.012

Xu, K., Mai, J., He, L., Yan, X., & Chen, Y. (2015). Surface Electromyography of Wrist Flexors and Extensors in Children With Hemiplegic Cerebral Palsy. PM and R, **7**(3), 270–275. doi: 10.1016/j.pmrj.2014.09.009.

Zhao, L., Dodge, T., Nemani, A., & Yokota, H. (2014). Resonance in the mouse tibia as a predictor of frequencies and locations of loading-induced bone formation. *Biomechanics and Modeling in Mechanobiology*, **13**(1), 141–151. doi: 10.1007/s10237-013-0491-2.