GENDER DIFFERENCE IN IMPACTS DURING RUNNING

JiSeon Ryu
Department of Health Care, Korea National Sport University, Seoul, Korea

The goal of this research was to determine whether gender differences exist in impact force and impact shock variables at stance phase during a preferred velocity running. Twelve male and ten female recreational runners were selected for this study. Comparisons of parameters relating to impact force and impact shock, which attained from time domain, and impact shock parameters, which were analyzed in frequency domain, were made between genders. There were no significantly differences in a magnitude of impact force and head and tibia acceleration between genders, but PSD (power spectral density) of peak impact shock at the tibia ($p <. 05$). Both male and female runners attenuated power between the tibia and head in 12-20 Hz impact range. However, female recreational runners exhibited that peak impact shock attenuation in the transfer function moved toward a high frequency band.

KEY WORDS: running, gender; impact force, impact shock

INTRODUCTION: During running, impact force and impact shock transmitted to the body. If these impacts work positively on the body, the musculature and bone would be strengthened. On the other hand, it has been known that repetitive impact, affecting on the joint and cartilage of the lower extremity, was one of major factors to occur injury and pain during running (Collins & Whittle, 1989). Impacts are being attenuated by muscle, cartilage and the action of the lower extremity's segment movement during the locomotion of human (Nigg et al., 1995; Shorten & Winslow, 1992). It was reported that female runners were vulnerable to running injuries, such as patellofemoral pain syndrome, iliotibial band friction syndrome, and tibial stress fractures compared to male runners (Taunton et al., 2002). It was speculated that gender difference in physical structure might affect on running mechanics, which may lead to specific running injuries. Little study has been devoted to differences in impact force, impact shock, and shock attenuation between male and female during running. The goal of this research was to determine whether gender differences exist in impact force and impact shock variables at stance phase during a preferred velocity running.

METHODS: Twelve male and ten female subjects volunteered to participate in this study (mean $\pm$ SD male height: 178.7 $\pm$ 5.5 cm, female height: 165.5 $\pm$ 4.3 cm; male mass: 72.8 $\pm$ 3.0 kg. female mass: 58.9 $\pm$ 3.2 kg; male age: 25.4 $\pm$ 4.8, female age: 23.5 $\pm$ 3.1). They were asked to run at their preferred running speed (male preferred running speed: 2.6 $\pm$ 0.2 m/s; female: 2.0 $\pm$ 0.1 m/s) with heel-toe running and step on a force plate (9286AA Model, Kistler, Switzerland) embedding on the ground. Axial accelerations of the tibia and mouth were measured using low-mass accelerometers (8776A50 Model, Kistler, Switzerland, 4.5 g). One accelerometer was fasten using surgical tape and attached onto the distal portion of the right tibia. The other attached to a long stick to bite with the mouth. Data from a force plate and both accelerometers were sampled with 1000 Hz for 20 sec. To take off the impact force from a vertical ground reaction force curve, a high-pass Butterworth filter of 4th order was used. In addition, a 4th order Bandpass filter was applied to separate impact shock wave out from acceleration signals. The power spectral densities (PSD) of two acceleration signals from each 3 strides were determined in FFT (fast Fourier transformation) techniques. Acceleration samples beyond toe-off were set to zero to remove non-contact phase data from the sample. Active peak and impact peak of the impact shock during the stance phase were found at 4 to 8 Hz and 12-20 Hz, respectively (Shorten and Winslow, 1992). The attenuation of the impact shock between the tibia and head was determined in the time domain using peak accelerations (PA):$IA = (1-PA_{mouth}/PA_{tibia})$. Attenuation was also calculated in the frequency domain using a transfer function equation: $T = 10\log_{10} (PSD_{mouth} / PSD_{tibia})$. Comparisons of parameters relating to impact force and impact shock attained from time domain, and impact shock parameters, which were analyzed in frequency domain, were made between genders. A repeated-measures ANOVA was used to test the difference of these variables between genders (p <. 05), but shock attenuation was assessed qualitatively.

RESULTS AND DISCUSSION: Table 1 summarized all of the impact force and shock variables between genders during a preferred velocity running. Figure1 showed average transfer function to assess the impact shock attenuation from the tibia up to the head.

Table1 Mean and SD of Impact Force and Shock for Male and Female Group.

Variables		Male	Female	F-value
Time Domain	Impact force peak (BW)	1.10 ± 0.31	0.74 ± 0.22	1.75
	Time to impact peak (ms)	25.8 ± 5.0	26.0 ± 5.4	0.25
	Tibia's peak acc. (g)	3.8 ± 0.8	2.5 ± 0.8	3.40
	Time to tibia's peak acc. (ms)	22.1 ± 6.1	25.1 ± 1.0	3.70
	Mouth's peak acc. (g)	0.82 ± 0.2	0.69 ± 0.3	0.60
	Shock attenuation	0.77 ± 0.13	0.73 ± 0.15	0.31
Frequency Domain	PSD of tibia's impact shock peak (g^2/Hz)	0.0221 ± 0.0019	0.0032 ± 0.0004	9.52*
	Tibia's shock peak frequency (Hz)	20	19	
	PSD of tibia's active peak (g^2/Hz)	0.0132 ± 0.0051	0.0279 ± 0.0007	2.81
	Tibia's active peak frequency (Hz)	7	4	
	PSD of mouth's max. shock peak (g^2/Hz)	0.0670 ± 0.0220	0.0334 ± 0.0181	0.65
	Mouth's max. peak frequency (Hz)	2	2	

* p < 0.05

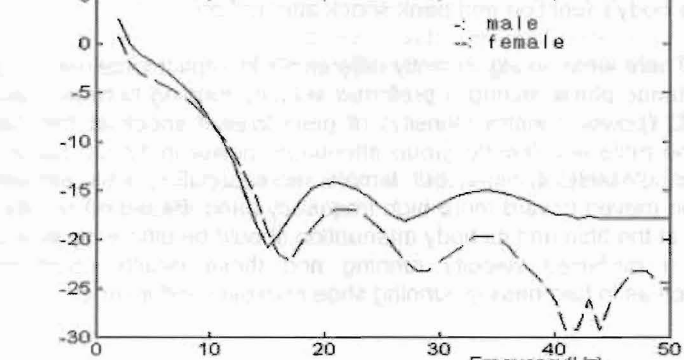

Figure 1 Transfer Functions between Tibia and Head Accelerations for each Gender.

Though it was reported that the gender difference in a musculoskeletal structure and kinematical movement of the body segments, playing a role in attenuating the impact during locomotion, little quantification has been conducted a difference in impact variables between genders during a preferred velocity running. The vertical peak impact force, generating harmful stress (Collins & Whittle, 1989), of the male group was greater by 0.36 BW than female group, having a lower preferred-running speed as it was suggested that vertical impact force increased with running speed (Nigg et al.1987; Frederick et al. 1981), but not statistically difference between genders.

Peak acceleration of the tibia in this study, which coincided with previous study, indicated that it was affected by running speed (Clarke et al., 1985; Lafortune & Hennig, 1988), but a significant difference was rarely made between genders. It was speculated that peak

acceleration was depended on the body tissue's damping effects for impact as well as running speed.
Peak passive shock of the tibia in frequency domain of the male group was significantly greater than that of the female group. It meant that the male group took relatively greater shock at the tibia compared to the female group during running regularly –it would be affected by gender difference in a preferred velocity for running.
It was important to notice the attenuation of impact shock from the tibia up to the head from clinical point of view, because repeated shock waves during running associated with degenerating the joints (Simon et al., 1972; Radin et al., 1973; Milgrom, 1989). The body attenuation of peak acceleration from time function was completed by 70 to 80% in both genders, which supported the previous findings (Hamill et al, 1995; Valiant et al; 1987; Shorten & Winslow, 1992), but didn't significantly show the difference between genders. The transfer function used to examine attenuation of spectral power from the tibia to the head showed that both the male and female group attenuated power in 12 -20 Hz, which might be implicated in musculoskeletal injury (Nigg et al., 1995; Radin et al, 1973). These results in both groups agreed with previous studies (Hamill et al, 1995; Shorten & Winslow, 1992; Voloshin et al, 1985). However, the peak shock attenuation in transfer function showed that the male group was around 15 Hz and female group moved toward higher frequency band. Voloshin et. Al. (1985) suggested that patient's, being recovery from the march fracture of a bone, peak shock attenuation moved toward a high frequency band at least by 10 Hz compared to healthy person. Based on this previous study, the hypothesis that female's body was relatively more vulnerable to impact shock than male at a preferred velocity running. It was also speculated that the difference of peak shock attenuation in transfer function between genders was resulted from the gender difference in muscle function against shock (Paul et al. 1978; Michael et al., 2003), structure and function of joints and segments (Yoshioka et al., 1989), and kinematical function of the lower extremity (Tiberio, 1987; Cowan et al., 1996; Mizuno et al., 2001) during running. Further work will be needed to examine the relationship these body's function and peak shock attenuation.

CONCLUSION: There were no significantly differences in impact force, mouth and tibia peak acceleration at stance phase during a preferred velocity running between genders, but the difference in PSD (power spectral density) of peak impact shock at the tibia was made ($p < .05$). Both the male and female group attenuated power in 12 -20 Hz, which might be implicated in musculoskeletal injury, but female recreational runners exhibited that peak impact attenuation moved toward more high frequency band. Based on results of this study, the impact shock at the tibia and its body attenuation should be differently evaluated between genders during a preferred velocity running and these results could be applied to consideration, such as in hardness of running shoe midsoles and inserts.

REFERENCES:

Clarke, T.E., Cooper, L.B., Clark, D.E. & Hamill, C.L. (1985). The effect of increased running speed upon peak shank deceleration during ground contact. *Biomechanics* IX-B (pp.101-105). Champaign, IL: Human Kinetics.
Collin's., Whittle, M.W. (1989). Impulsive forces during walking and their clinical implications, *Clinical Biomechanics 4*, 179-187.
Cowan, D.N., Jones, B.H., Frykman, P.N., Polly Jr., D.W., Harman, E.A., Rosenstein, R.M. & Rosenstein, M.T. (1996). Lower limb morphology and risk overuse injury among male infantry trainees. *Med. Sci. Sports Exerc.* 28, 945-952.
Frederick, E.C., Hagy, J.L. & Mann, R.A. (1981). The prediction of the vertical impact force during running [Abstract]. *J. of Biomechanics, 14*, 498.
Hamill, J., Derrick, T., R. & Holt, K., G. (1995). Shock attenuation and stride frequency during running. *Human Movement Science, 14*, 45-60.
Lafortune, M.A. & Hennig, E.M. (1988). Effects of velocity and uphill slope on tibial shock during running. In C.E. Cotton, M. Lamontagne, D.G.E. Robertson & J.P. Stothart (Eds.), *Proceedings of the Fifth Biennial Conference and Human Locomotion Symposium of the*

Canadian Society of Biomechanics (pp.94-95). Ottawa: University of Ottawa, Dep. of Kinanthropology.

Michael, J., Decker, Michael R. Torry, Douglas J. Wyland, William I. Sterett, J. Richard Steadman. (2003). Gender differences in lower extremity kinematics, kinetics and energy absorption during landing. *Clinical Biomechanics* 18, 662-669.

Milgrom, C., Giladi, M., Kashtan, H., Simkin, A., Chisin, R., Margulies, J., Steinberg, R., Aharonson, Z. & Stein, M. (1985). A prospective study of the effect of a shock-absorbing orthotic device on the incidence of stress fractures in military recruits. *Foot and Ankle,* 6, 101-104.

Mizuno, Y., Kumagai, M., Mattessich, S.M., Elias, J.J., Ramrattan, N., Cosgarea, A.J. & Chao, E.Y. (2001). Q-angle influences tibiofemoral and patellofemoral kinematics. *J. Orthop. Res.* 19, 834-840.

Nigg, B.M., Bahlsen, H.A., Luethi, S.M. & Stokes, S. (1987). The influence of running velocity and midsole hardness on external impact forces in heel-toe running. *J. of Biomechanics*, 20, 951-959.

Nigg and Cole. (1995). Impact forces during heel-toe running. *J. Biomech.*11: 407-432.

Paul I.L., Munro M.B, Abernethy, S.S.R, Radin E.L. & Rose R.M. (1978). Musculo-skeletal shock absorption: Relative contribution of bone and sort tissue at various frequencies. *J. of Biomechanics,* 11, 237-239.

Radin, E.L., Paul, I.L. & Rose, R.M. (1972). Role of mechanical factors in pathogenesis of primary osteoarthritis. *Lancet*, 519-522.

Simon, S.R., Radin, E.L. & Paul, I.L. (1972). The response of joints to impact loadings, Part II: In vivo behavior of subchondral bone. *J. of Biomechanics,* 5, 267-272.

Shorten, M., R. & Winslow, D., S. (1992). Spectral analysis of impact shock during running. *International Journal of Sport Biomechanics, 8,* 288-304.

Taunton, J.E., Ryan, M.B., Clement, D.B., McKenzie, D.C., Lloydsmith, D.R. & Zumbo, B.D. (2002). A retrospective case-control analysis of 2002 running injuries. *Br. J. Sports Med.*36, 95-101.

Tiberio, D. (1987). The effect of excessive subtalar joint pronation on patellofemoral mechanics: a theoretical model. *J. Orthop. Sports Phys. Ther.*9, 160-165.

Voloshin A.S., Burger C.P., Wosk J. & Arcan M. (1985). An in vivo Evaluation of the leg's shock absorbing capacity. In: B. Winter et. al. (Ed), *BiomechanicsIX*-B (112-116). Champaign, IL: Human Kinetics.

Yoshioka, Y., Siu, D.W., Scudamore, R.A. & Cooke, T.D. (1989). Tibial anatomy and functional axes. *J. Orthop. Res.*7, 132-137.