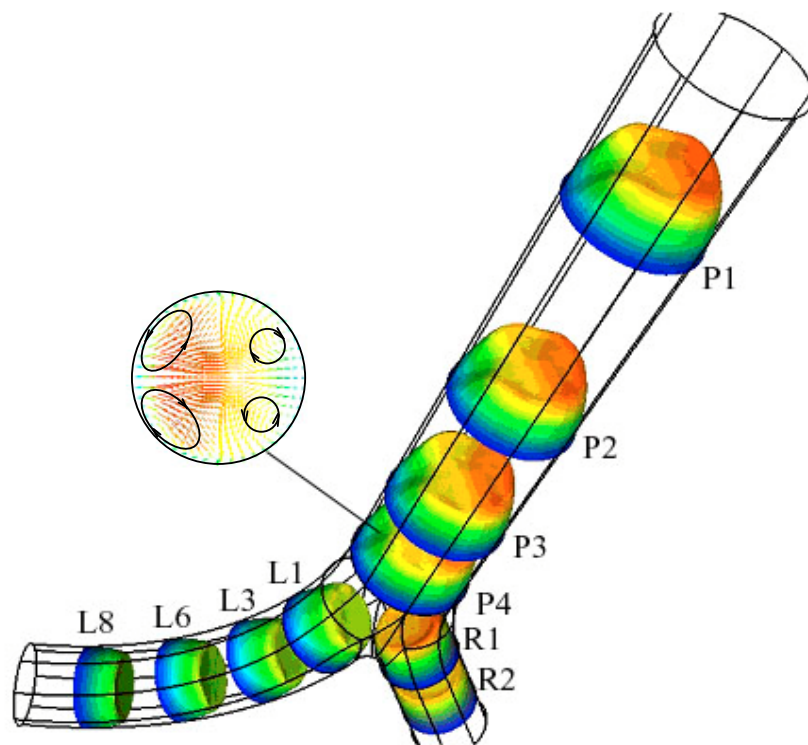


4.2.2 จังหวะการหายใจออก (Expiration phase)

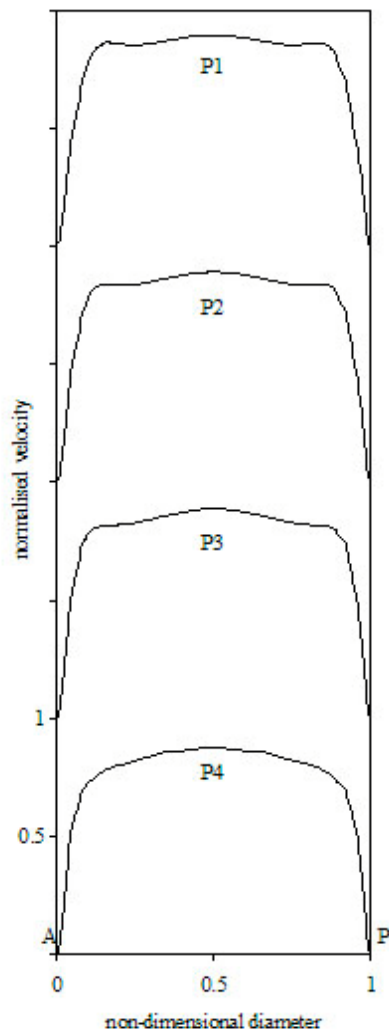
ในช่วงจังหวะการหายใจออกภายใต้สภาวะการหายใจขณะออกกำลังกาย จะมีการกำหนดความเร็วที่ทางเข้า Daughter airways ทั้งซ้ายและขวา โดยความเร็วที่กำหนดจะมีลักษณะเป็น uniform velocity profile โดยที่มีขนาดของความเร็วที่เป็นฟังก์ชันที่ขึ้นกับเวลาตาม sinusoidal ฟังก์ชัน อัตราการไหลที่ใช้กำหนดที่ทางเข้า daughter airways ซ้ายและขวาจะเป็นสัดส่วน 45% และ 55% ของอัตราการไหลที่ trachea ตามลำดับ ความเร็วสูงสุดในช่วงจังหวะการหายใจออกที่วัดได้ที่ trachea จะให้ค่า Reynolds number (Re) = 4.66×10^4 ซึ่งจะมีค่าเท่ากับความเร็วสูงสุดในช่วงจังหวะการหายใจเข้า ซึ่งการกำหนดค่า numerical boundary condition ต่างๆ ดังกล่าวจะมีการกำหนดเหมือนกับสภาวะการหายใจแบบปกติ

จากผลที่ได้จากการวิเคราะห์การไหลของอากาศด้วยโปรแกรม CFD จะพบว่าเมื่อมีการไหลจากท่อทางเดินหายใจซ้ายและขวา อากาศจะไหลเข้ามารวมกันในท่อหลัก parent airway หลังจากที่มีการรวมกันของอากาศที่ไหลจาก daughter airway ทั้งสองที่ bifurcation จะเกิดการไหลวนของอนุภาคของไหล ซึ่งจะทำให้เกิด secondary flow 2 คู่ที่ parent airway ซึ่งแสดงไว้ในรูปของเวกเตอร์ความเร็วที่หน้าตัดต่างๆ ใน parent airway ดังแสดงใน รูปที่ 18 การเกิด secondary flow ใน parent airway นี้เนื่องมาจากการเปลี่ยนทิศทางการไหลอย่างกะทันหันของอากาศที่ไหลมาจาก daughter airways ซึ่งจะปรากฏ secondary flow อย่างชัดเจนตรงจุด bifurcation และ secondary flow จะเริ่มหายไปเมื่อการไหลมีการพัฒนาตัวผ่านท่อตรงของ parent airway ซึ่งจะสังเกตได้จาก velocity profile ที่จุดต่างๆ ใน parent airway ตามแนว R-L ในรูปที่ 18 จะเห็นว่าลักษณะของ velocity profile จะเปลี่ยนจาก profile ที่มีจุดเยื้องไปทางขวาที่จุด P4 (ใกล้ bifurcation) ไปเป็น profile ที่คล้ายพาราโบลาที่จุด P1 (ระยะห่างจาก bifurcation ออกไปตามทิศทางการไหล)

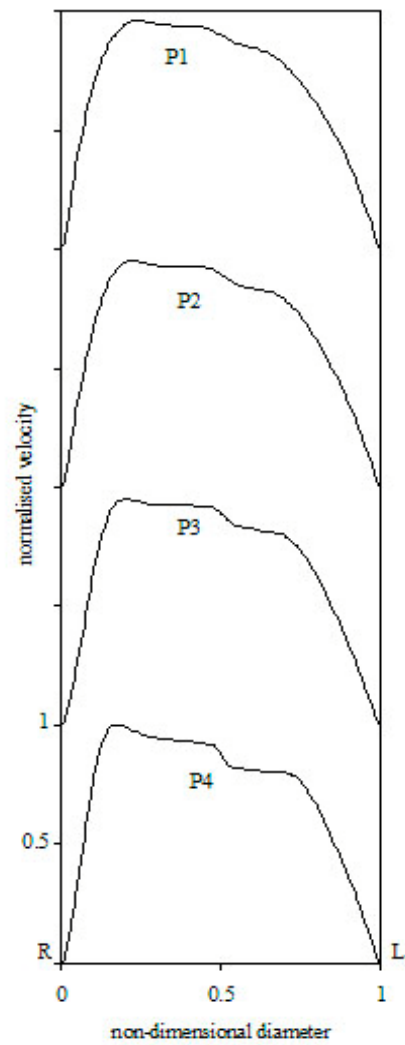
รูปที่ 19 จะแสดงให้เห็นในรูปของการไหลของอนุภาคของไหลดังแสดงใน ซึ่งจะเห็นว่าอนุภาคของของไหลที่อยู่ติดกับผนังด้านนอกของ bifurcation จะเกิดการเปลี่ยนทิศทางมากที่สุดทำให้เกิด secondary flow ของอนุภาคของของไหลที่ผนังด้านนอกของ bifurcation แต่หลังจากที่อากาศไหลผ่านท่อตรงของ parent airway เนื่องจากอิทธิพลของความหนืดทำให้การไหลพัฒนาไปสู่ fully developed และทำให้ secondary flow หายไป



รูปที่ 17 แสดงลักษณะการไหลที่หน้าตัดต่างๆที่ความเร็วสูงสุด ($Re = 4.66 \times 10^4$) ในช่วงจังหวะหายใจออก (Expiration cycle)

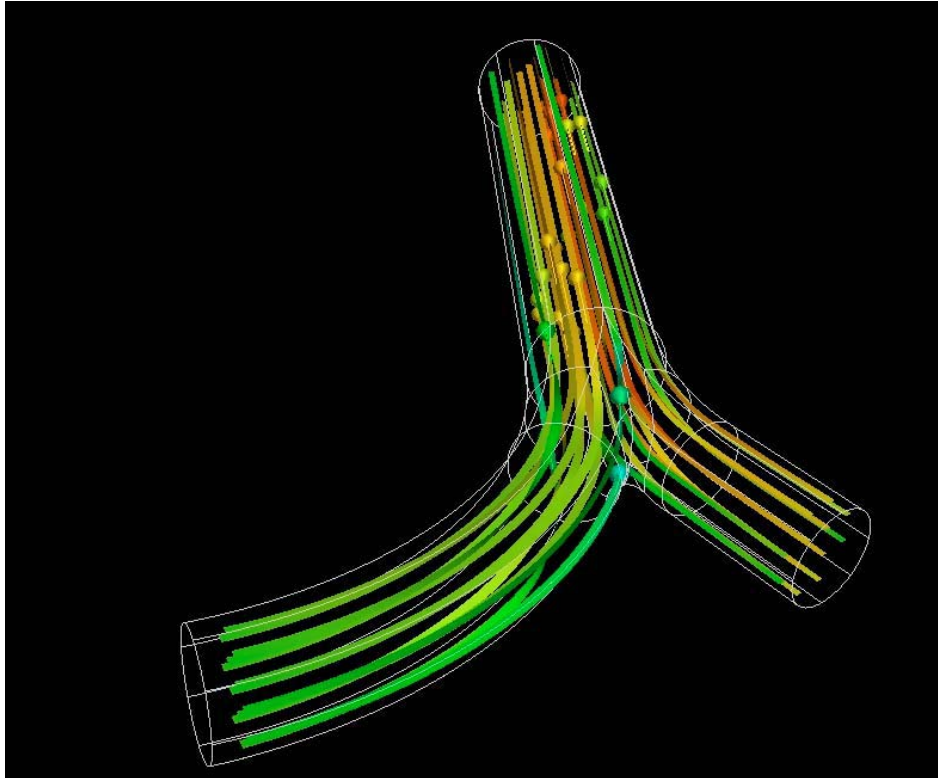


a) Parent airway (A-P)



b) Parent airway (R-L)

รูปที่ 18 แสดง **velocity profile** ที่หน้าตัดต่างในท่อทางเดินหายใจด้านขวา (a) และซ้าย (b) ที่ความเร็วสูงสุดในช่วงจังหวะการหายใจออกภายใต้สภาวะการหายใจแบบ **maximal exercise condition** ($V_T = 3.33$ L, $Re = 4.66 \times 10^4$ and $\alpha = 4.747$)



รูปที่ 19 ลักษณะการไหลของอนุภาคของไหลในจังหวะหายใจเข้าที่ความเร็วสุด ($Re=4.66 \times 10^4$) ที่ทางออก trachea ภายใต้สภาวะการหายใจแบบ maximal exercise condition

4.3 การหายใจที่ความถี่สูง High-Frequency Ventilation Condition (HFV)

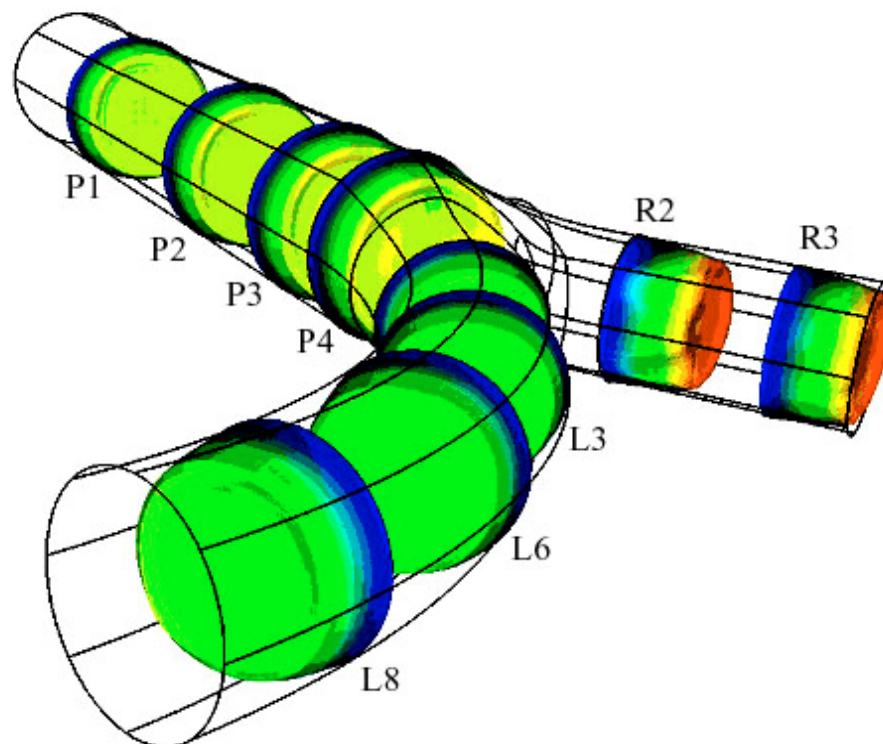
สภาวะการหายใจแบบ HFV condition จะมีค่าตัวแปรต่างๆ ที่ใช้กำหนดค่าในการคำนวณ ดังนี้

- Tidal volume (V_T) = 0.05 L
- Breathing frequency (f) = 5 Hz
- Time period (T) = 0.2 sec/cycle
- Time increment (Δt) = 0.02 sec
- Maximum mean velocity at trachea (U_{max}) = 3,906.25 mm/s
- Reynolds number (Re) = 4.37×10^3
- Wormesley number (α) = 11.87
- Large Eddy Simulation (LES) model

4.3.1 จังหวะการหายใจเข้า (Inspiration phase)

สำหรับลักษณะของรูปแบบการไหลภายใต้สภาวะ HFV จะแตกต่างจาก resting condition และ maximal exercise condition ค่อนข้างมาก โดยจากรูปที่ 20 จะเห็นว่า ลักษณะของ velocity profile ของของไหลที่แต่ละหน้าตัดเริ่มตั้งแต่ทางเข้า parent airway ลงมาตลอดท่อทางเดินหายใจ จนถึง bifurcation ลักษณะของ velocity profile จะค่อนข้างเป็นรูปแบบ flat profile ซึ่งจะมีลักษณะใกล้เคียงกับ uniformed velocity profile ที่กำหนดที่ทางเข้า และความหนาของ boundary layer ก็มีความหนาไม่เปลี่ยนแปลง ลักษณะการไหลแบบนี้ยังเป็นไปในรูปแบบเดียวกันเมื่ออากาศไหลผ่าน bifurcation ไปยัง daughter airway ทั้งสอง โดยที่ velocity profile จะมีลักษณะคล้าย uniformed velocity profile ที่มียอดที่ติดกับผนังด้านในของ bifurcation สูงกว่าทางผนังด้านนอก ดังแสดงในรูปที่ 21 นอกจากนั้นยังไม่พบว่ามี secondary flow เกิดขึ้นดังปรากฏใน 2 กรณีที่กล่าวมาแล้ว

จะสังเกตว่าลักษณะการไหลของอากาศภายใต้สภาวะการหายใจแบบ HFV ซึ่งเป็นการไหลแบบ oscillatory ภายใต้ความถี่ค่อนข้างสูง จะมีลักษณะของ profile ค่อนข้างที่จะรักษาสภาพเหมือน profile ของความเร็วที่กำหนดทางด้านขาเข้า parent airway และความหนาของ boundary layer ค่อนข้างที่จะคงที่ตามทิศการไหล นั้นแสดงว่าอิทธิพลเนื่องจากความหนืดจะมีผลต่อการไหลค่อนข้างน้อย และเนื่องจากความถี่ของการไหลแบบ oscillatory ค่อนข้างสูงดังนั้นช่วงเวลาที่การไหลจะพัฒนาตัวให้อยู่ภายใต้อิทธิพลของความหนืดจึงมีค่อนข้างน้อย ซึ่งจะเห็นว่าลักษณะของการไหลที่ได้จากคอมพิวเตอร์วิเคราะห์การไหลในงานวิจัยนี้ จะคล้ายกับงานที่ได้จากการศึกษาจากการทดลองที่ Re สูงสุดของ Nishida et al (1997)



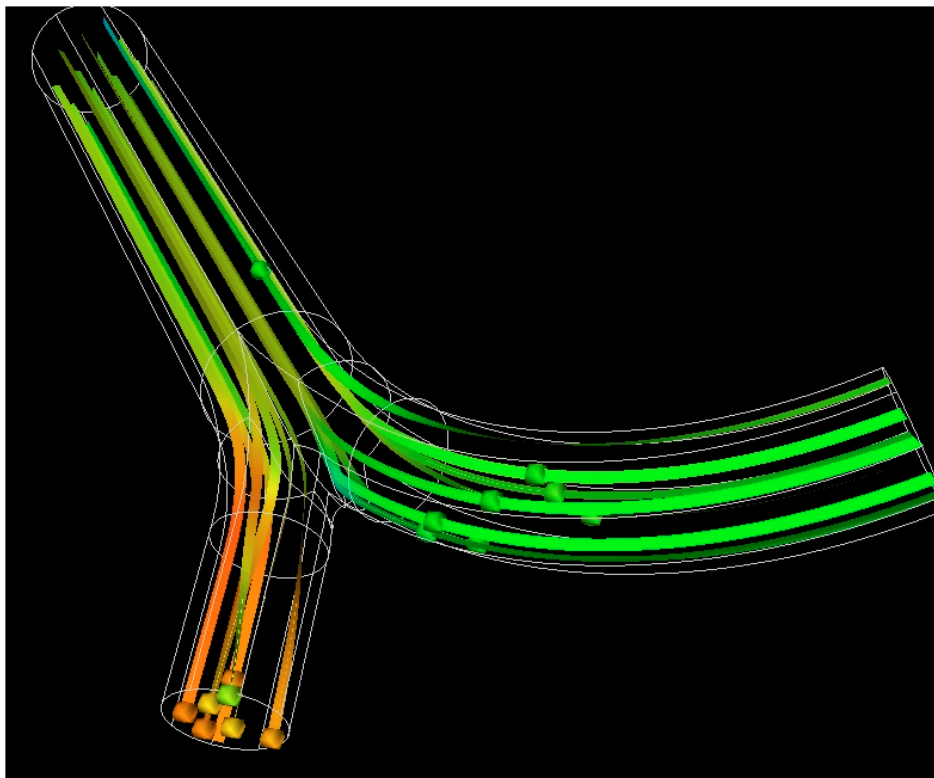
รูปที่ 20 แสดงลักษณะการไหลที่หน้าตัดต่างๆ ที่ความเร็วสูงสุด ($Re = 4.37 \times 10^3$) ในช่วงจังหวะหายใจเข้า (Inspiration cycle)



a) Right daughter airway

b) Left daughter airway

รูปที่ 21 แสดง velocity profile ที่หน้าตัดต่างในท่อทางเดินหายใจด้านขวา (a) และซ้าย (b) ที่ความเร็วสูงสุดในช่วงจังหวะการหายใจเข้าภายใต้สภาวะการหายใจแบบ HFV condition ($V_T = 0.05$ L, $Re = 4.37 \times 10^3$ and $\alpha = 11.87$)

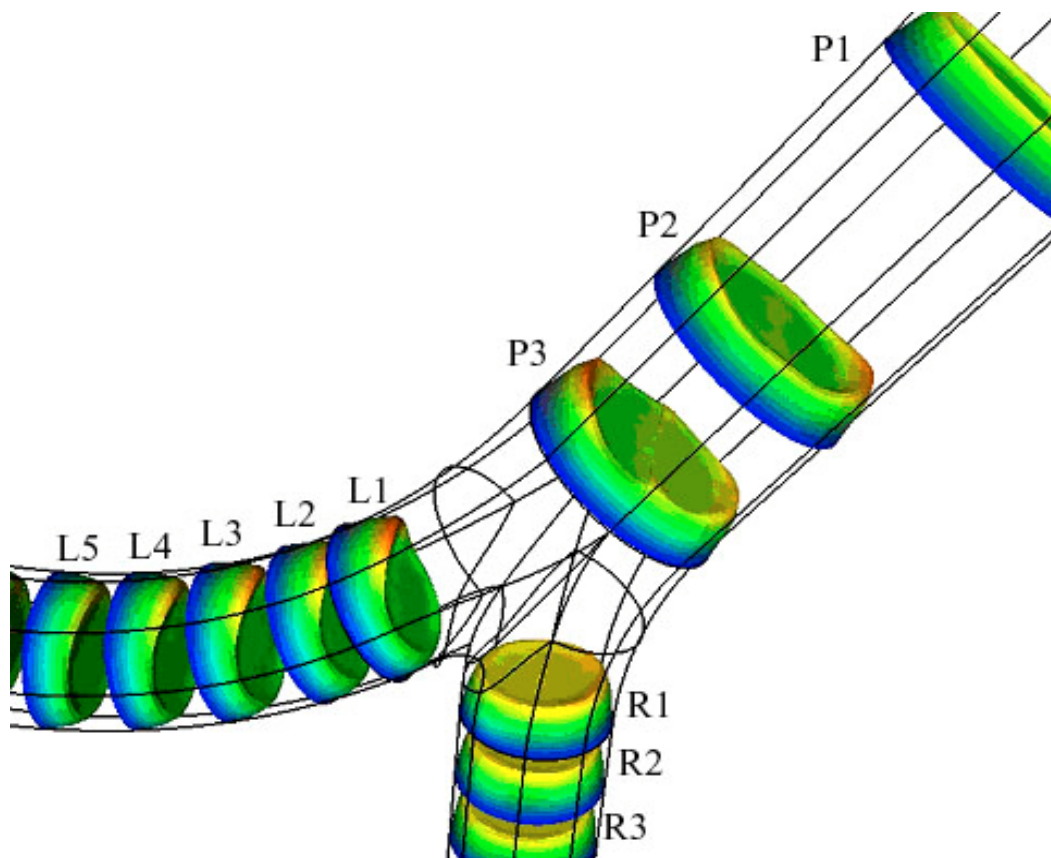


รูปที่ 22 ลักษณะการไหลของอนุภาคของไหลในจังหวะหายใจเข้าที่ความเร็วสุด ($Re=4.37 \times 10^3$) ที่ทางเข้า trachea ภายใต้สภาวะการหายใจแบบ HFV condition

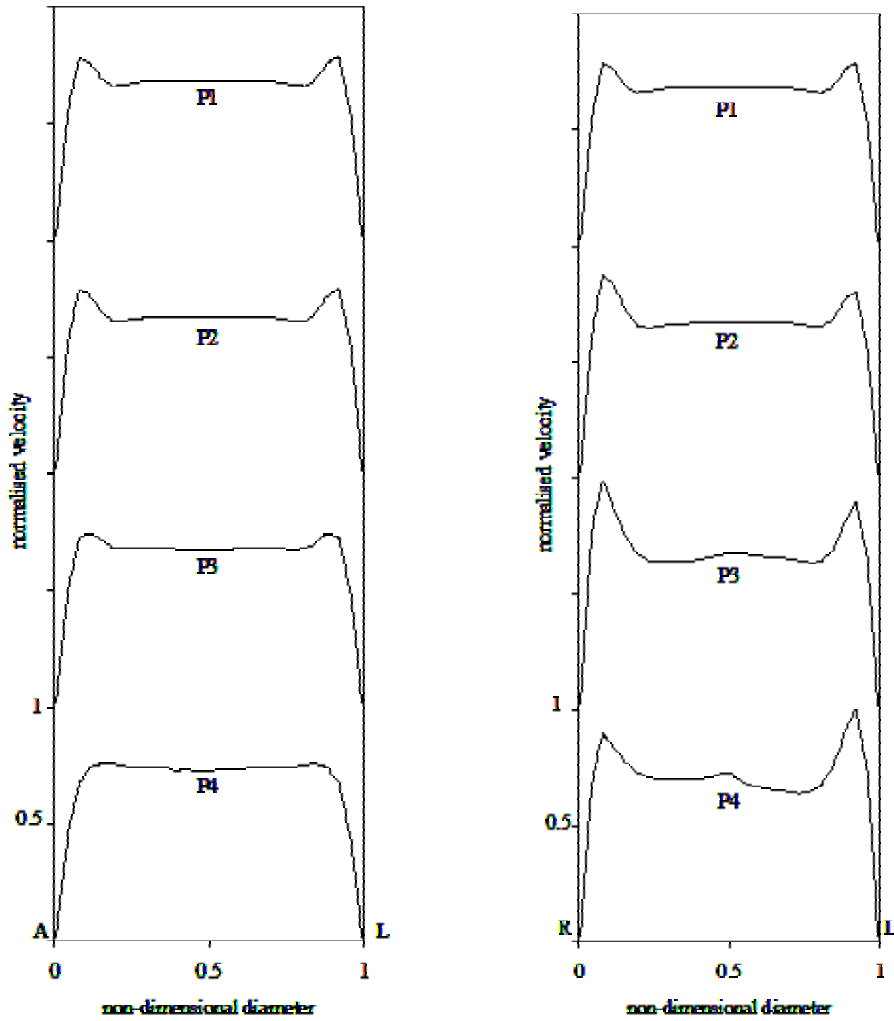
4.3.2 จังหวะการหายใจออก (Expiration phase)

ลักษณะการไหลของอากาศภายในท่อระบบทางเดินหายใจภายใต้สภาวะ HFV ในช่วงของการหายใจออกจะมีลักษณะคล้ายกับการหายใจเข้าภายใต้สภาวะเดียวกัน นั่นคือการไหลจะไม่ขึ้นอยู่กับอิทธิพลของความหนืด แต่จะขึ้นอยู่กับเวลาและลักษณะของ entrance flow profile และรูปแบบการไหลตามทิศทางการไหลจะมีลักษณะคล้ายกับ profile ที่ทางเข้า ดังแสดงในรูปที่ 23 ซึ่งจะเห็นว่า entrance flow profile มีลักษณะเป็น uniformed velocity profile ที่ทางเข้า daughter airway ด้านซ้ายและขวา หลังจากให้อากาศไหลผ่านท่อทางเดินลักษณะของ flow profile ยังคงมีลักษณะคล้ายกับ entrance flow profile จนกระทั่งการไหลผ่านส่วนโค้งของท่อทางเดิน ซึ่งจะเห็นว่า profile ของของไหลที่อยู่ติดกับด้านในของส่วนโค้งจะมียอดยกสูงขึ้นมากกว่าด้านนอก เนื่องจากความเร็วของอนุภาคของของไหลที่ติดกับผนังด้านในของส่วนโค้งมีความเร็วมากกว่าด้านนอก

เมื่อลำการไหลจาก daughter airway ทั้งสองไหลมารวมกันที่ bifurcation และไหลเข้าสู่ parent airway ลักษณะของ velocity profile จะมีลักษณะเป็นยอดตัดทางด้านขวาที่สูงกว่าทางด้านซ้าย เนื่องจาก flow rate ที่มาจาก daughter airway ด้านขวามากกว่าทางด้านซ้ายและลักษณะของ flow profile ก็เป็นนี้ไปเรื่อยๆ จนกระทั่งสิ้นสุดที่ทางออก parent airway นอกจากนั้นยังไม่พบ secondary flow สำหรับการหายใจภายใต้สภาวะ HFV และที่ช่วงเวลาอื่นๆ ในจังหวะการหายใจออก ลักษณะของการไหลก็จะมีลักษณะเดียวกันกับที่อธิบายมาข้างต้น



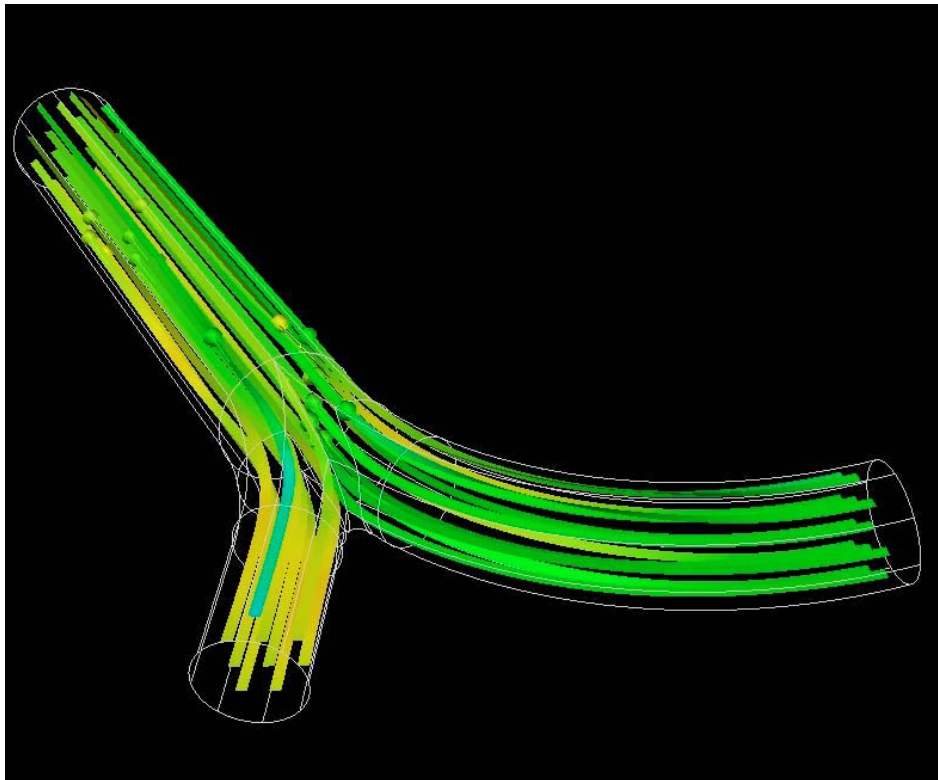
รูปที่ 23 แสดงลักษณะการไหลที่หน้าตัดต่างๆ ที่ความเร็วสูงสุด ($Re = 4.37 \times 10^3$) ในช่วงจังหวะหายใจออก (Expiration cycle)



a) Parent airway (A-P)

b) Parent airway (R-L)

รูปที่ 24 แสดง velocity profile ที่หน้าตัดต่างในท่อทางเดินหายใจด้านขวา (a) และซ้าย (b) ที่ความเร็วสูงสุดในช่วงจังหวะการหายใจออกภายใต้สภาวะการหายใจแบบ HFV condition ($V_T = 0.05$ L, $Re = 4.37 \times 10^3$ and $\alpha = 11.87$)



รูปที่ 25 ลักษณะการไหลของอนุภาคของไหลในจังหวะหายใจเข้าที่ความเร็วสุด ($Re=4.37 \times 10^3$) ที่ทางออก trachea ภายใต้สภาวะการหายใจแบบ HFV condition

5 บทวิจารณ์

จากการศึกษาลักษณะการไหลของอากาศภายในระบบทางเดินหายใจในส่วนของ conducting airway ซึ่งจะเริ่มตั้งแต่ trachea และ bronchus ข้ายและขวา ซึ่งจะเป็นส่วนที่จะศึกษาในงานวิจัยนี้ โดยจะศึกษาภายใต้สภาวะการหายใจแบบต่าง ๆ ได้แก่ การหายใจที่สภาวะปกติ (Resting condition) การหายใจภายใต้สภาวะการออกกำลังกาย (Maximal exercise condition) และ การหายใจภายใต้การหายใจที่มีความถี่สูง (High-frequency ventilation, HFV) การศึกษาในงานวิจัยนี้จะใช้โปรแกรมการวิเคราะห์การไหลเชิงตัวเลข (Computational Fluid Dynamic, CFD) สำหรับศึกษาการไหลของอากาศภายในโครงสร้างของระบบท่อทางเดินหายใจแบบ 3 มิติ ที่มีลักษณะเป็นแบบท่อยกที่ไม่สมมาตร (Asymmetric single bifurcation) ซึ่งจะมีรูปร่างและขนาดใกล้เคียงกับของจริงมากที่สุด (Horsefield et al, 1973) และการกำหนดเงื่อนไขของการไหลในงานวิจัยนี้จะจำลองรูปแบบของการหายใจด้วยการไหลแบบ Oscillatory flow ซึ่งเป็นการไหลที่ขึ้นกับเวลาและใกล้เคียงกับลักษณะของการหายใจจริง ซึ่งจะต่างจากงานวิจัยอื่นๆที่มักจะศึกษาการไหลดังกล่าวภายใต้สภาวะ steady state ในท่อทางเดินหายใจที่เป็นท่อยกแบบสมมาตร (Zhao and Lieber, 1994a,b; Gatlin et al, 1995; Gatlin et al, 1997)

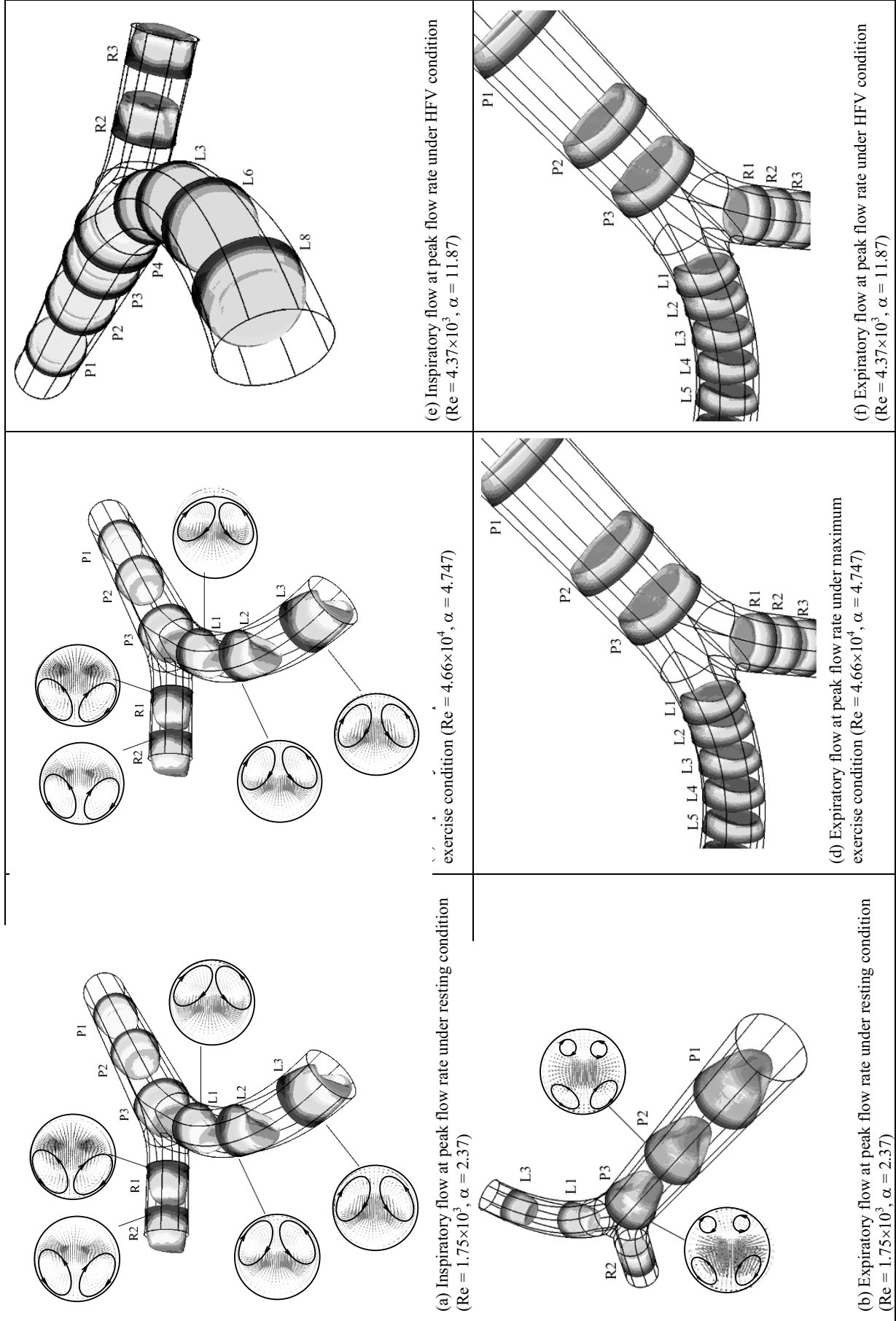
จากการศึกษารูปแบบของการไหลของอากาศในระบบท่อทางเดินหายใจแบบ asymmetric single bifurcation ในงานวิจัยนี้พบว่า สำหรับสภาวะการหายใจแบบปกติ ($V_T = 0.5 \text{ L}$, $f = 0.2 \text{ Hz}$, $Re = 1.75 \times 10^3$) และสภาวะการหายใจขณะออกกำลังกาย ($V_T = 3.33 \text{ L}$, $f = 0.8 \text{ Hz}$, $Re = 4.66 \times 10^4$) รูปแบบการไหลของอากาศจะมีลักษณะคล้ายกัน ทั้งในช่วงจังหวะการหายใจเข้าและหายใจออก จะต่างกันเล็กน้อยในช่วงจังหวะการหายใจเข้า ซึ่งจะพบการเกิด secondary flow เพิ่มอีก 1 คู่ที่ carinal ridge (จุดแยกตัวของลำการไหลที่ bifurcation) ในขณะที่รูปแบบการไหลที่สภาวะปกติจะไม่พบ secondary flow ที่ carinal ridge แต่จะพบที่ผนังด้านนอกของ bifurcation เท่านั้น แต่อย่างไรก็ตามเมื่ออากาศไหลผ่านส่วนที่เป็น carinal ridge ไปแล้ว secondary flow ที่ carinal ridge จะหายไป จะพบเพียง secondary flow ที่เกิดที่ผนังด้านนอกของ bifurcation เท่านั้น ซึ่งจะเป็นไปในรูปแบบเดียวกันสำหรับทั้งสองกรณี ดังแสดงใน รูปที่ 26 (a) และ (c) จากลักษณะรูปแบบการไหลที่เกิดขึ้นจะเห็นว่าการไหลของอากาศในระบบท่อทางเดินหายใจภายใต้การหายใจแบบปกติและขณะออกกำลังกายจะมีอิทธิพลเนื่องมาจากความหนืดของของไหล ซึ่งจะสังเกตได้จากความหนาของ boundary layer ที่เพิ่มขึ้นตามทิศทางการไหล และลักษณะของ velocity profile ที่พยายามพัฒนาตัวให้เป็นรูปแบบ parabolic profile แต่เนื่องจากความยาวในท่อทางเดินหายใจในแต่ละช่วงจะมีส่วนที่เป็นท่อดังที่มีความยาวไม่มากพอที่จะทำให้ของไหลเกิดการพัฒนาตัวไปเป็น fully develop

flow ทำให้ entrance flow effect ยังคงมีผลต่อลำการไหลตามทิศทางการไหล จากการศึกษาค้นคว้ารูปแบบของการไหลของอากาศภายใต้สภาวะการหายใจที่มีความถี่สูง หรือ HFV condition ($V_T = 0.05 \text{ L}$, $f = 5 \text{ Hz}$, $Re = 4.37 \times 10^3$) จะให้ผลที่แตกต่างจากการหายใจแบบปกติและขณะออกกำลังกายเป็นอย่างมาก โดยจะเห็นว่าจะไม่เกิด secondary flow ในการไหลสภาวะนี้ และรูปแบบของการ flow profile จะขึ้นกับรูปร่างของ entrance flow อย่างเดียว ดังแสดงใน รูปที่ 26 (e) และ (f) จะเห็นว่า การไหลภายใต้สภาวะ HFV จะเป็นการไหลแบบ oscillatory flow ภายใต้ความถี่การหายใจที่ค่อนข้างสูงดังจึงมีเวลาไม่มากพอในการพัฒนารูปแบบของการไหลให้เป็นการไหลแบบ fully develop จะเห็นว่าความหนืดจะมีอิทธิพลต่อการไหลค่อนข้างน้อยต่อการไหลแบบ HFV ซึ่งจะสังเกตได้จากความหนาของชั้น boundary layer ยังมีค่าไม่เปลี่ยนแปลงแม้การไหลจะมีการไหลผ่านทางเดินหายใจก็ตาม

จากรูปแบบการไหลของอากาศในระบบทางเดินหายใจที่สภาวะการหายใจแบบต่างๆ จะพบว่า มีประโยชน์ต่อการพิจารณาการพ่นยาเข้าสู่ระบบทางเดินหายใจ นั่นคือจะพบว่า ละอองยาจะไปติดและสะสมอยู่ในบริเวณที่เกิด secondary flow เช่น ในกรณีการหายใจแบบปกติจะพบว่า เมื่อมีการพ่นละอองยาพร้อมกับการหายใจเข้า ละอองยาจะมีการสะสมอยู่ที่ผนังด้านนอกของส่วนที่เป็น bifurcation แต่จะไม่พบการสะสมที่ carinal ridge และสำหรับกรณีการหายใจภายใต้สภาวะที่มีการหายใจเอาปริมาณอากาศเข้าไปมากๆ เช่น การออกกำลังกาย จะพบว่า ละอองยาจะมีการสะสมมากขึ้นทั้งทางผนังด้านนอกของ bifurcation และที่ carinal ridge ซึ่งจะแตกต่างจากการหายใจแบบความถี่สูง (HFV) ที่ไม่พบ secondary flow ในลำการไหลเลยนั้น แสดงว่า เมื่อมีการพ่นละอองยาเข้าไปกับลำการไหลภายใต้สภาวะการหายใจแบบนี้ จะเกิดการผสมของละอองยาตามผนังท่อทางเดินหายใจน้อยมาก ทำให้ตัวปริมาณยาที่จะเข้าไปสู่ alveoli เพื่อที่เข้าสู่กระแสเลือดมีปริมาณมากกว่าการหายใจทั้งสองแบบข้างต้น

อย่างไรก็ตาม การศึกษารูปแบบการไหลของอากาศในระบบทางเดินหายใจโดยการจำลองการไหลของอากาศในงานวิจัยนี้ยังมีข้อจำกัดอยู่หลายอย่างในเรื่องการพิจารณา และกำหนดสภาวะการไหล ดังนั้น เพื่อที่จะทำให้ได้การไหลของอากาศในระบบทางเดินหายใจที่สมจริงมากที่สุด จะต้องมีการพิจารณาเรื่องของรูปแบบของ entrance flow profile ที่มีผลต่อลักษณะการไหลของอากาศเมื่อมีการไหลผ่านท่อลงไป ซึ่งการจะกำหนดรูปแบบของ entrance flow profile ให้สมจริงนั้นเป็นไปได้ค่อนข้างยาก นอกจากนั้นแล้ว การจำลองการไหลที่มีการไหลผ่านท่อที่มีสารหล่อลื่นเคลือบก็เป็นส่วนที่จะต้องนำมาพิจารณา เนื่องจากที่ผนังท่อทางเดินหายใจจะมีเมือก (mucus) เคลือบอยู่ อีกทั้งเนื้อเยื่อของท่อทางเดินหายใจที่มีความยืดหยุ่น และการถ่ายเทความร้อนระหว่างร่างกายกับอากาศที่ไหลเข้า ก็เป็นส่วนที่จะต้องนำมาพิจารณาต่อไป เพื่อให้ได้รูปแบบการไหลที่สมจริงมากที่สุด อีกทั้งข้อจำกัดในเรื่อง เวลาในการประมวลผลของคอมพิวเตอร์ ขนาดของหน่วยความจำของคอมพิวเตอร์ และตัวโปรแกรมการวิเคราะห์การไหลที่มีอยู่ ทำให้ไม่สามารถที่จะ

กำหนดสภาวะการไหลให้สมจริงทั้งหมดได้ ดังนั้นตัวโปรแกรมการวิเคราะห์การไหลจึงจำเป็นที่จะต้องมีการพัฒนาต่อเนื่องเพื่อที่สามารถที่จะทำให้สามารถคำนวณการไหลที่มีลักษณะสมการทางคณิตศาสตร์ที่มีความซับซ้อน และเป็นระบบที่ใหญ่ ภายใต้โครงสร้างที่ซับซ้อนแบบ 3 มิติ ของระบบทางเดินหายใจ



รูปที่ 26 รูปแสดงการเปรียบเทียบรูปแบบการไหลภายใต้สภาวะการหายใจแบบต่าง ๆ

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ภาคผนวก

Numerical simulation of respiratory airflow at high-frequency ventilation (HFV) condition within the human upper airway

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Abstract

This study represents numerical investigations of respiratory airflow (or oscillatory flow) mechanism in a three-dimensional asymmetric single bifurcation model of the human central airway based on Horsfield's morphology data (Horsfield et al, 1971). Two different breathing conditions, normal breathing condition and high-frequency ventilation (HFV) condition, were studied utilising a finite volume CFD code. Peak Reynolds number (Re) of 1.75×10^3 and Womersley number (α) of 2.37 were used for the normal breathing condition, this correspond to the tidal volume (VT) and the breathing frequency (f) of 0.5L and 0.2Hz, respectively. For the HFV condition, the peak Reynolds number of 4.37×10^3 and Womersley number (α) of 11.87 were used and corresponding to the tidal volume (VT) and the breathing frequency (f) of 0.05L and 5Hz, respectively. The results showed that secondary motions are negligible for the HFV condition while the secondary motions are found very strong for the normal breathing condition. The respiratory flow results for the normal breathing condition are found strongly dependent on the convective effect and the viscous effect with some contribution of the unsteadiness effect. Whereas the respiratory flow under the HFV condition is only influenced by the unsteadiness effect.

Introduction

The respiratory diseases such as asthma, emphysema and bronchitis are connected with air pollutions in the nowadays environment and the number of these diseases tends to increase. The therapy of such diseases often uses pharmaceuticals, which are directly administered in the form of aerosols or dispersed particles to the respiratory system. The efficacy and the effect of those pharmaceuticals are dependent on particle transport, size of particles and the deposition of particles in the lung. However, very little information on the pattern of the particle deposition or the effectiveness of the aerosol treatments is available, and this forms an effective area of clinical respiratory research. Knowledge of the airflow mechanism within the airways system is the first step in the understanding of the movement of the particles and their deposition in the human respiratory airways network.

Methods

An asymmetric single bifurcation model of the central airway based on the morphological data "Model 1:zone1" given by Horsfield et al (1971) was used in the present study to simulate the respiratory airflow under HFV conditions. This model represents the central airway including the trachea, right and left lobar bronchi as shown in **Error! Reference source not found.** The details of airway lengths, diameters, branching angle and the

ratio of curvature radius (R) to daughter airway diameter (d) together with the airflow rate distributions in each branch are given in Table 1.

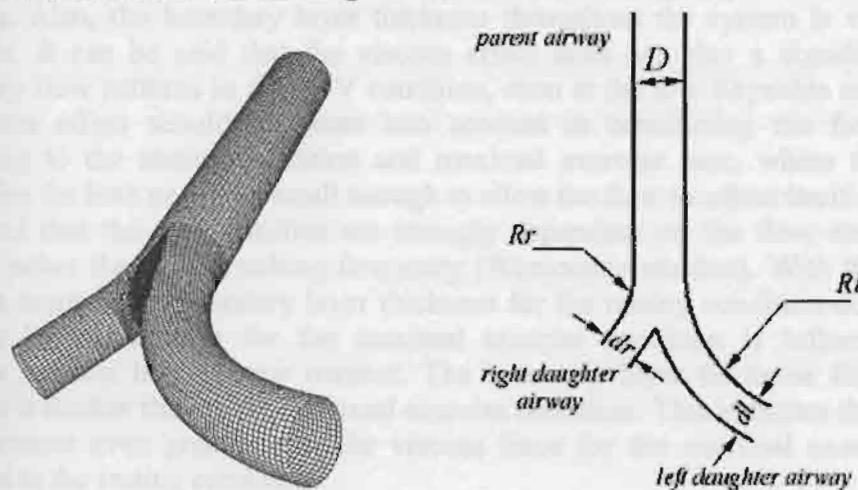


Table 1 Geometric details and flow distribution in the central airway

Branch No.	Diameter (mm)	Length (mm)	Branching angle ($^{\circ}$)	Axial curvature (R)	R/d	Flow (% of trac
0 (parent branch)	$D=16$	100	-	-	-	100.00
1 (left daughter branch)	$d_l=12$	50	73°	54	4.5	45.0342
10 (right daughter branch)	$d_r=11.1$	22	35°	33.5	3.0	54.9658

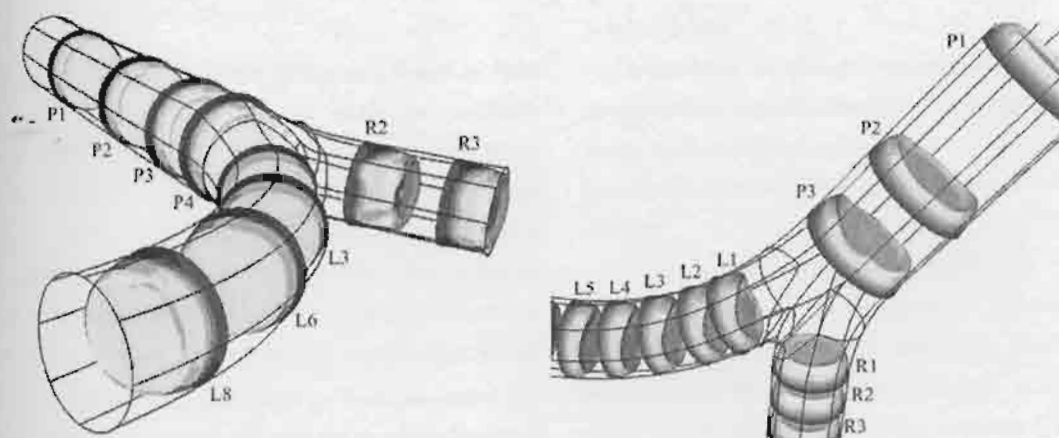
Note: Dimensions given here pertain to 75% TLC and each branch has identification corresponding to that given by (Horsfield et al., 1971)

Due to the airway geometrical complexity, technique of multi block method was applied to ease the geometry and mesh generation. The airway mesh model consists of 84 blocks with structured hexahedral mesh cells of 159,872 node points. At the regions closed to the airway walls and the carinal ridge, the mesh cells were refined with well mesh grading ratio in order to capture high velocity gradients occurring near the airway walls and the carinal ridge. The Navier-Stokes equations are solved using a CFD code with the concurrent block Jacobi (CBJ) algorithm. The CBJ code is based a finite volume method with second order upwind scheme for convective numerical fluxes and central differencing for the diffusive numerical fluxes. The implicit scheme is used within each interior domain. In the CFD context, second order accuracy is sufficient to give the accurate results. The implicit method is also used for solution stability thus limitation of time increment can be avoided.

Results

For the HFV condition, the axial velocity profiles in the parent tube on expiration are significantly different from the profiles found in the resting condition and maximal exercise condition. The velocity profiles in the parent airway for the HFV condition are rather flat in the core of the tube with higher velocity near the both side of the tube. When

the flow develops downstream, this higher peak velocity remains near the wall, but with less amplitude. Secondary motions are not found during expiration for the HFV condition. Also, the boundary layer thickness throughout the system is very thin and consistent. It can be said that the viscous effect does not play a significant role on respiratory flow patterns in the HFV condition, even at the low Reynolds number where the viscous effect should be taken into account in considering the flow behavior. Comparing to the resting condition and maximal exercise case, where the breathing frequencies for both cases are small enough to allow the flow to adjust itself to develop, it was found that the flow profiles are strongly dependent on the flow rate (Reynolds number) rather than the breathing frequency (Womersley number). With the increasing Reynolds number the boundary layer thickness for the resting condition decreases. The boundary layer thickness for the maximal exercise condition is influenced by the Reynolds number in the same manner. The boundary layer thickness for the resting condition is thicker than in the maximal exercise condition. This indicates that the inertia force becomes even greater than the viscous force for the maximal exercise case as compared to the resting condition.



Conclusion

It can be concluded that Reynolds number is the important parameter influencing the respiratory flow as long as the breathing frequency is small enough to allow the flow to adjust itself to develop. This also suggests that the flow in the low breathing frequency range is affected by viscous effect, thus the geometry of the airway is of important. In other words, the respiratory flow is dependent on the Dean number when the Womersley number is small (small breathing frequency). However, the viscous effect becomes less important when the breathing frequency becomes high, such as high-frequency ventilation. Thus the geometry effect can be neglected for the HFV condition and only the frequency parameter (Womersley number) plays a significant role on the respiratory flow patterns.

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Simulation of the Unsteady Oscillatory Flow in Three-dimensional Asymmetric Bifurcation Model of the Conducting Airway in the Human Lungs

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Abstract

A computational fluid dynamics (CFD) code based on finite volume method (FVM) was used to study the unsteady respiratory airflow dynamics within the human lung. The three-dimensional asymmetric single bifurcation model of the central airway based on the morphological data given by Horsfield et al was used in the present study to simulate the oscillatory respiratory flow. Numerical simulation was made for the resting or normal breathing condition. Peak Reynolds number (Re) number of 1.75×10^3 and Womersley number (α) of 2.73 were used, this corresponds to the tidal volume (VT) and the breathing frequency of 0.5L and 0.8 Hz, respectively.

The simulation of respiratory flow within the single bifurcation was found to be sufficient to give a number of results which both qualitatively and quantitatively agree well with other published experimental works ([1], [3], [9], [10], [11], [12], [14] and [15]). The respiratory flow results were found strongly dependent on the convective effect and the viscous effect with some contribution of the unsteadiness effect. The secondary motions were observed at the outer walls of the branching curve of two daughter airways during inspiration, and in left daughter airway, and parent airway during expiration.

1. Introduction

Modeling of the airflow in the conducting airway in the human lungs is important to predict the particle deposition of both contaminate and the pharmaceuticals. In order to achieve good predictions of the deposition the airflow has to be modeled correctly.

Because of the complexity of the bronchial tree structure, the use of the small airway and highly asymmetric bifurcation structure, *in-vivo* studies are difficult for obtaining the global airways effect and the detail of the smaller airways downstream. While the experimental studies are crucial to obtain a somewhat satisfied detail by utilizing physiological scale-up model to study the airflow dynamics in the lungs. However those experimental studies are limited up to the 3rd generation of bifurcation and the smaller airways are in accessible to the most experimental technique even in the scale-up model ([1], [3], [11] and [12]).

Computational fluid dynamics (CFD) studies through the solutions of the Navier-Stoke equations seem to be a better way to access the smaller airways and easy to show a large number of data in three dimensional physiologically realistic airway geometries. Together with accurate computer modeling it should be possible to develop the insight to understand airflow pattern and the particle transport that can provide information which will avoid inappropriate clinical trials and waste effort.

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Mesh Model and Construction

The asymmetric single bifurcation model of the central airway based on the morphological data given by Horsfield et al [1] was used in the present study to simulate the respiratory flow under resting condition.

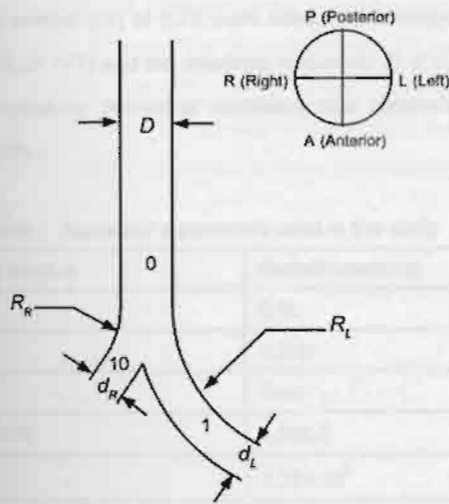


Figure 1 Model of single bifurcation of central airway (trachea and two main right and left bronchi).

Table 1 Model description of a single bifurcation model [1]

Branch No.	Diameter (mm)	Length (mm)	Branching angle	R/d	Axial curvature (R)
0 (parent branch)	D=16	100	-	-	-
1 (left daughter branch)	d ₁ =12	50	$\beta=73^\circ$	4.5	54
2 (right daughter branch)	d ₂ =11.1	22	$\alpha=35^\circ$	3.0	33.5



Figure 2 Three-dimensional mesh model of an asymmetric single bifurcation of the central airway

Due to the airway geometrical complexity, technique of multi block method was applied to ease the geometry and mesh

generation. The airway mesh model consists of 84 blocks with structured hexahedral mesh cells of 159,872 node points. At the regions closed to the airway walls and the carinal ridge, the mesh cells were refined with well mesh grading ratio in order to capture high velocity gradients occurring near the airway walls and the carinal ridge.

The airway geometry dimensions and mesh model are given in Table 1, Figure 1 and Figure 2.

3. Respiratory Flow

The respiratory flow can be assumed to be incompressible laminar flow. The Navier-Stokes equations of the time-dependent incompressible flow are expressed by

Continuity equation:

$$\nabla \cdot \mathbf{V} = 0 \quad (1)$$

Momentum equations:

$$\text{x-component } \frac{\partial u}{\partial t} + (\mathbf{V} \cdot \nabla)u = -\frac{1}{\rho} \frac{\partial p}{\partial x} + \nu \Delta u \quad (2-a)$$

$$\text{y-component } \frac{\partial v}{\partial t} + (\mathbf{V} \cdot \nabla)v = -\frac{1}{\rho} \frac{\partial p}{\partial y} + \nu \Delta v \quad (2-b)$$

$$\text{z-component } \frac{\partial w}{\partial t} + (\mathbf{V} \cdot \nabla)w = -\frac{1}{\rho} \frac{\partial p}{\partial z} + \nu \Delta w \quad (2-c)$$

4. Numerical Boundary Condition and Solution

The Navier-Stokes equations are solved using a CFD code based FVM with the concurrent block Jacobi algorithm. The implicit scheme is used with the second order upwind scheme for the convective numerical flux and the central differencing for the diffusive numerical flux.

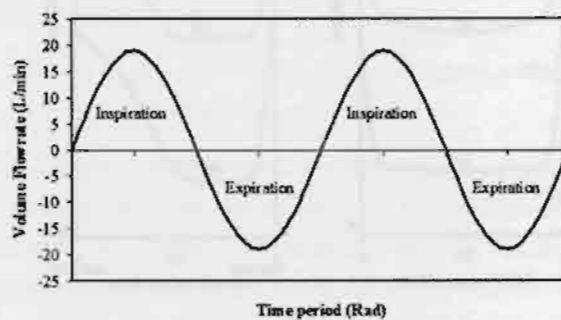


Figure 3 Respiratory cycle at resting condition

The airway walls can be assumed solid wall, hence no-slip boundary condition can be applied to the airway walls. To regulate oscillatory flow and control the flow rates in right and left daughter airways, the velocity boundary condition was applied on the inflow/outflow boundaries in the airway. The controlled flow

the right and left daughter airways are 55% and 45% of the peak velocity in the trachea, respectively. The uniform velocity boundary conditions applied to the inflow/outflow boundary vary with time as a sinusoidal time function as shown in Figure 3.

Peak Reynolds number (Re) number of 1.75×10^3 and Womersley number (α) of 2.73 were used, this corresponds to a tidal volume (VT) and the breathing frequency (f) of 0.5L and 10 Hz, respectively. Numerical respiratory flow parameters are given in Table 2

Table 2 Numerical parameters used in the study

Breathing condition	Normal breathing
VT (L)	0.5L
f (Hz)	0.2Hz
T (sec)	5sec
$\bar{U}_{Tmax}$ (mm/s)	1,562.5
Re	1.75×10^3
Time steps /cycle	100
Time increment, Δt	0.05sec
Dimensionless time at each time increment, $t^* = 4 \nu \Delta t / D^2$	0.01

5. Results and Discussions

5.1 The respiratory flow results

The results are presented as the velocity profiles normalized by the maximum average local velocity in each branch. The results are taken from 8 cross-sections station throughout the model. The oscillatory flow simulation results in a three-dimensional asymmetric bifurcation model at resting condition case are given in two separate parts, inspiration and expiration.

5.1.1 Inspiratory flow

An overview of the flow features for the inspiration phase under the resting condition is illustrated in Figure 4. During inspiration, uniform inlet was applied at the parent airway. The velocity profile at station P1 is still uniform and it develops downstream through a straight pipe passing station P2 and P3 of the parent airway with thicker boundary layer near the wall. When the flow has developed further downstream the peak velocity is found just next to the boundary layer near the wall as shown in Figure 4. The peak velocity near the wall is due to the diffusion of momentum over the cross-section while the flow tends to adjust itself during the oscillatory period. However this peak velocity is less noticeable further downstream and the velocity profile develops to the parabolic profile characteristic. When the flow from parent tube reaches the carinal ridge the flow has not yet

fully developed, although the airway length is about 6.25 local diameters. This is because the time increment that allowed the velocity to change in magnitude is not large enough to allow the flow to adjust itself to be fully developed.

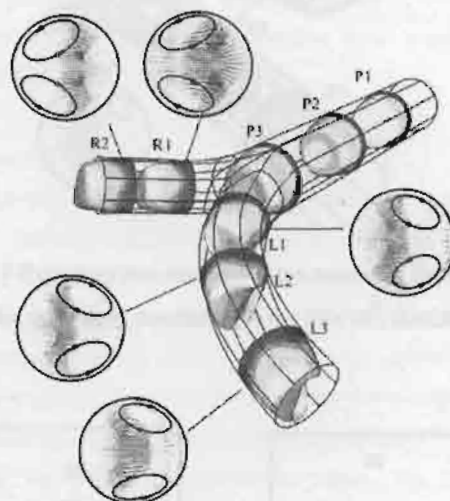


Figure 4 Inspiratory flow features at the maximum inspiration under the resting condition ($Re = 1.75 \times 10^3$, $\alpha = 2.37$)

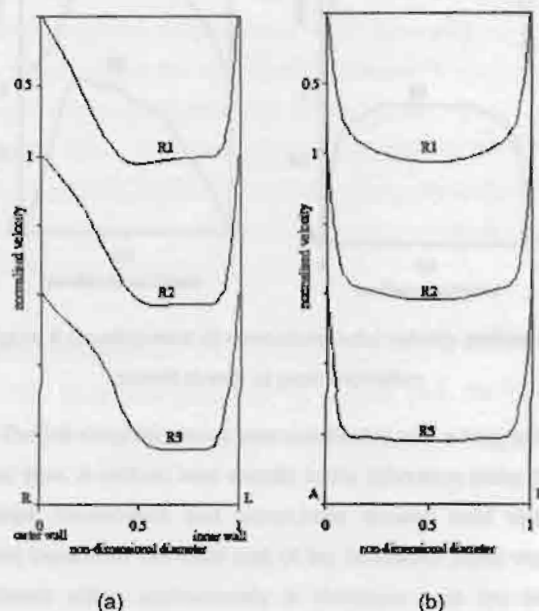


Figure 5 Development of normalized axial velocity profiles in right daughter airway at peak inspiration

At the carina the incompletely fully developed flow from the parent airway is split into two daughter airways. When the flow enters two daughter branches, the axial velocities in bough daughter airways in the bifurcation plane (R-L) are initially

flowed towards to the outer walls of the bifurcation (the inner side of bend). The maximum velocity profiles are found next to the inner wall of the bifurcation as shown in Figure 5a and Figure 5b. While for the normal plane to the bifurcation (A-P), symmetric velocity profiles are found in both daughter airways as shown in Figure 5b and Figure 6b. The profiles are initially parabolic and then develop to the "M-shape" or "bi-peak" velocity profiles. The distortion of the axial velocity profiles in both right and left daughter airways are due to the secondary motions occurring when the flow undergoes the curved portions of the bifurcation.

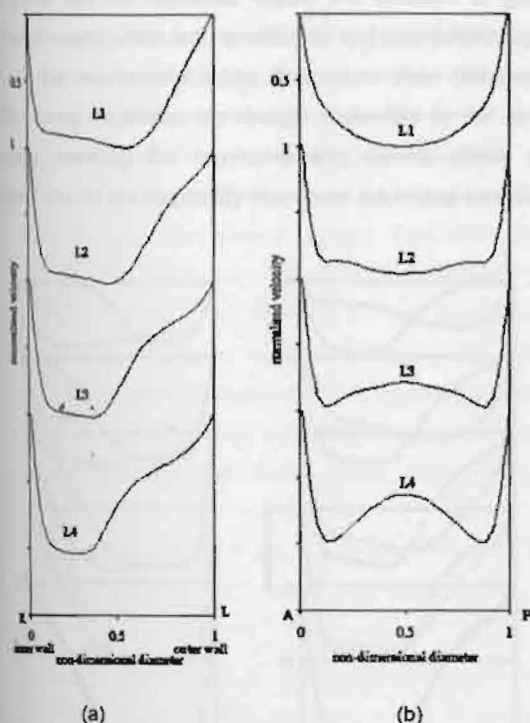


Figure 6 Development of normalized axial velocity profiles in left daughter airway at peak inspiration

5.1.2 Expiratory flow

In the expiration, the respiratory flow was regulated from the two daughter airways in which the uniform velocity profiles were imposed at the inflow boundaries. The expiratory flow feature is illustrated in Figure 7.

The right daughter is constructed with a short pipe continuing from the curved part at the bifurcation and the carinal ridge. When the flow moves through the straight pipe of right daughter airway at R2 and R1 velocity profile is still flat with a thicker boundary layer. The flow has not yet fully developed when it enters the bifurcation and the secondary effect was not found in right daughter airway during expiration phase.

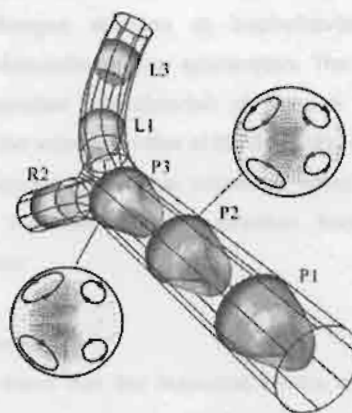


Figure 7 Expiratory flow features at the maximum Expiration under the resting condition ($Re = 1.75 \times 10^3$, $\alpha = 2.37$)

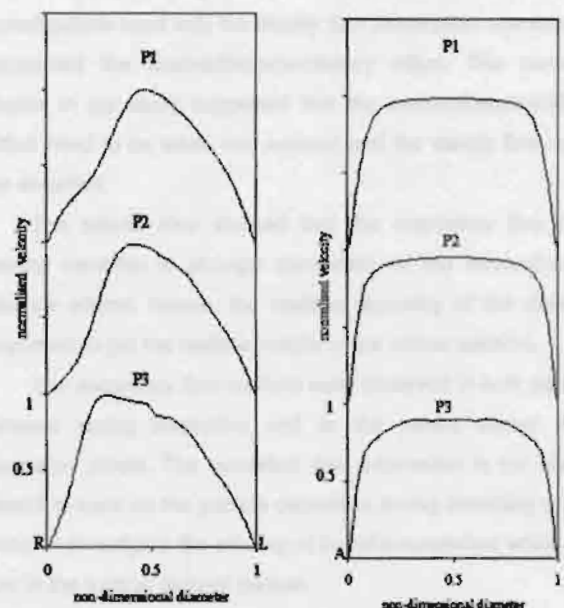


Figure 8 Development of normalized axial velocity profiles in parent airway at peak expiration

The left daughter airway was constructed with a long uniform curved pipe. A uniform inlet velocity in the bifurcation plane (R-L) develops downstream and pronounces skewed axial velocity profiles towards to the inner wall of the bifurcation (outer wall of the bend) within approximately 3 diameters from the inflow boundary before the flow reaches the bifurcation. The distortion of axial velocity profile in the left daughter airway during expiration is purely affected by the secondary flow caused by the curvature portion.

When two flow streams from the both daughter airways merge together in the bifurcation, two pairs of unequal vortices are formed downstream just above the carinal ridge in the parent

airway as shown in Figure 7. In the plane of bifurcation (R-L) peak velocity profile is found near the right wall because the flow rate regulated from the right daughter airway is greater than that driven from the left daughter airway. While the axial velocity profile in the normal plane to the bifurcation (A-P) is symmetric characteristic with incomplete fully developed profile.

5.3 Discussions on the oscillatory flow effects

The numerical results of inspiratory flow and expiratory flow at peak flow rate under the resting condition were compared to the experimental results given by [1], [3], [9], [11], [14] and [15]. It was found that the numerical results are sufficient to give a number of results which both qualitatively and quantitatively agree well with the experimental works. The results show that velocity profiles during inspiration are strongly dependent on the airway geometry, meaning the convective and viscous effects play important role on the respiratory flow under the resting condition.

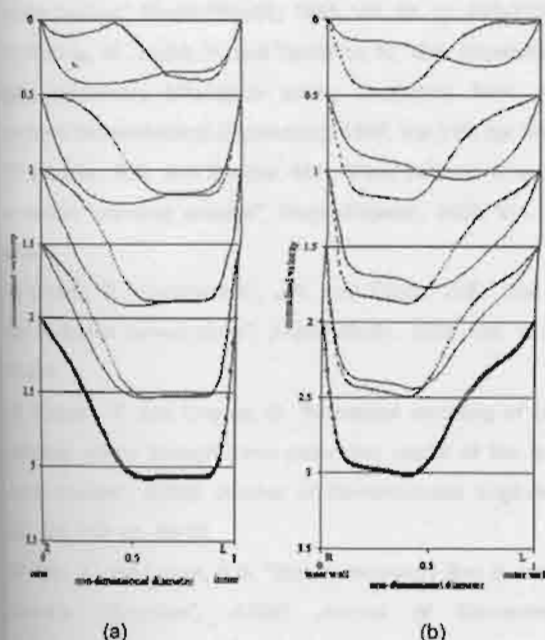


Figure 9 Axial velocity profiles in the bifurcation plane (R-L) during inspiration cycle at (a) station R2 in right daughter airway and (b) station L3 in left daughter airway

It is noted that, both inspiratory and expiratory flows results are agreed with the previous experimental results when the flow reaches the peak flow during respiratory cycle. However, at the other flow rates during the same respiratory cycle (but between acceleration and deceleration) the velocity profiles obtained from acceleration and deceleration are different as shown in Figure 9.

The flow separation at the near "zero" flow was observed when the flow changes direction at inspiration-to-expiration and expiration-to-inspiration during acceleration. The velocity profiles during acceleration are distorted at the low flow rate. The distortion of the velocity profiles at the beginning of the inspiration can be influenced by the flow separation effect at the reversal flow when the flow changes direction from expiratory to inspiratory flow.

6. Conclusions

It was found that the numerical results of the respiratory flow, both inspiratory and expiratory flows, in an asymmetric single bifurcation of the central airway at peak flow rate are sufficient to give a number of results which both qualitatively and quantitatively agree well with the experimental results given by [1], [3], [9], [11], [14] and [15]. However, those experimental investigations used only the steady flow assumption and were not concerned the unsteadiness/oscillatory effect. The numerical results in our study suggested that the unsteadiness/oscillatory effect need to be taken into account and the steady flow cannot be assumed.

The results also showed that the respiratory flow under resting condition is strongly dependent on the convective and viscous effects. Hence, the realistic geometry of the airway is important to get the realistic results of the airflow patterns.

The secondary flow motions were observed in both daughter airways during inspiration and in the parent airway during expiration phase. The numerical flow information in the study is useful to track on the particle deposition during breathing cycle in order to investigate the efficacy of the pharmaceutical which most are in the form of aerosol particle.

7. Acknowledgement

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การจำลองการไหลของอากาศในระบบทางเดินหายใจภายใต้สภาวะ HFV Numerical simulation of respiratory flow under HFV condition in the conducting airway

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Abstract

The unsteady respiratory airflow dynamics within the human tracheobronchial network under high-frequency ventilation (HFV) condition was investigated utilizing Computational Fluid Dynamics (CFD) technique based on a finite volume method (FVM). Peak Reynolds number (Re) number of 4.37×10^3 and Womersley number (α) of 11.87 were used, this corresponds to the tidal volume (VT) and the breathing frequency (f) of 0.05L and 5Hz, respectively. A three-dimensional single bifurcation model of the upper airway based on morphological model proposed by Horsfield et al. (1971) has been used. The simulation results for both inspiration and expiration phases agreed with experimental results given by Nishida et al (1997). The results were compared to our previous studies of the respiratory airflow simulations under resting ($Re = 1.75 \times 10^3$, $VT = 0.5L$ and $f = 0.2Hz$) and maximal exercise ($Re = 4.66 \times 10^4$, $VT = 3.33L$ and $f = 0.8Hz$) conditions [Calay et al. (2001)]. It was found that during the high-frequency ventilation (HFV) the flow is dominated by the unsteadiness effect, while the respiratory flows at resting condition and maximal exercise condition are dominated by the convective effect. It also was found that the respiratory flow under the high-frequency ventilation behaves in the similar way as in the straight tube. Hence the geometry effect becomes less important in such flow condition.

Keywords: Respiratory flow, HFV condition, Bifurcation

1. Introduction

The therapy of the respiratory diseases mostly uses the pharmaceuticals which are always in the form of aerosol delivered directly into the lungs. The efficiency and efficacy of the

therapy are dependent on the transport and deposition of drug particles and their sizes. The knowledge of respiratory airflow patterns and its mechanism is the first step in order to achieve the good understanding of the movement of the particles and their deposition in the respiratory airway network. However, there are very little investigations of the respiratory airflow within realistic respiratory airway geometry under realistic breathing conditions.

Because of the complexity of the bronchial tree structure, the use of the small airway and highly asymmetric bifurcation structure, *in-vivo* studies are difficult for obtaining the global airways effect and the detail of the smaller airways downstream. While the experimental studies are crucial to obtain a somewhat satisfied detail by utilizing physiological scale-up model to study the airflow dynamics in the lungs. However those experimental studies are limited up to the 3rd generation of bifurcation and the smaller airways are in accessible to the most experimental technique even in the scale-up model ([1], [7], [15], [16], [18] and [19]).

Computational fluid dynamics (CFD) studies through the solutions of the Navier-Stroke equations seem to be a better way to access the smaller airways and easy to show a large number of data in three dimensional physiologically realistic airway geometries. Together with accurate computer modeling it should be possible to develop the insight to understand airflow pattern and the particle transport that can provide information which will avoid inappropriate clinical trials and waste effort.

There also have been number of respiratory airflow dynamics studies using CFD technique in both two- and three-dimensional bifurcating tube models ([3], [4], [5] and [17]). However most respiratory airflow dynamics investigations are

based on quasi-steady or time dependent/oscillatory flow assumptions under normal breathing condition. Our previous investigation [2] of CFD simulations of respiratory flow under resting condition and maximal exercise condition showed that the different breathing conditions yield different respiratory flow patterns. For the normal breathing condition and maximal exercise condition, the breathing frequency is considerably low (0.2 Hz). In order to investigate the effect of the higher breathing frequency on the respiratory flow simulation the respiratory airflow dynamics under high-frequency ventilation (HFV) condition was studied within a three-dimensional asymmetric single bifurcation model using CFD method.

2. Mesh Model and Construction

The asymmetric single bifurcation model of the central airway based on the morphological data given by Horsfield et al [6] was used in the present study to simulate the respiratory airflow under resting condition.

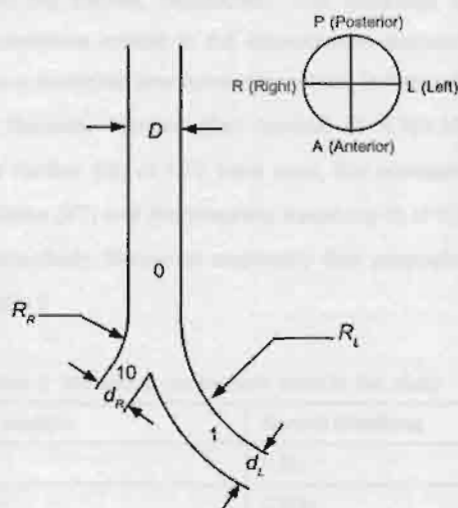


Figure 1 Model of single bifurcation of central airway (trachea and two main right and left bronchi).

Table 1 Model description of a single bifurcation model [6]

Branch No.	Diameter (mm)	Length (mm)	Branching angle	R/d	Axial curvature (R)
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1 (left daughter branch)	$d_L=12$	50	$\beta=73^\circ$	4.5	54
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Figure 2 Three-dimensional mesh model of an asymmetric single bifurcation of the central airway

Due to the airway geometrical complexity, technique of multi block method was applied to ease the geometry and mesh generation. The airway mesh model consists of 84 blocks with structured hexahedral mesh cells of 159,872 node points. At the regions closed to the airway walls and the carinal ridge, the mesh cells were refined with well mesh grading ratio in order to capture high velocity gradients occurring near the airway walls and the carinal ridge.

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3. Respiratory Flow

The respiratory flow can be assumed to be incompressible laminar flow. The Navier-Stokes equations of the time-dependent incompressible flow are expressed by

Continuity equation:

$$\vec{\nabla} \cdot \mathbf{V} = 0 \quad (1)$$

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4. Numerical Boundary Condition and Solution

The Navier-Stokes equations are solved using a CFD code based FVM with the concurrent block Jacobi algorithm. The implicit scheme is used with the second order upwind scheme for the convective numerical flux and the central differencing for the diffusive numerical flux.

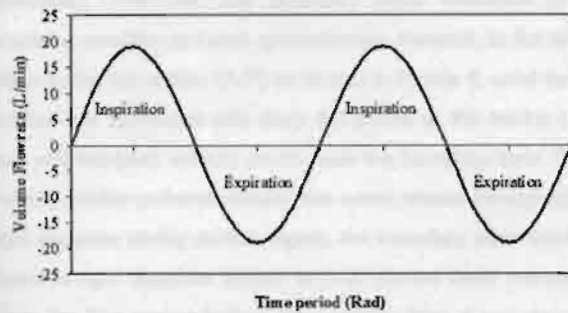


Figure 3 Respiratory cycle at resting condition

The airway walls can be assumed solid wall, hence no-slip boundary condition can be applied to the airway walls. To regulate oscillatory flow and control the flow rates in right and left daughter airways, the velocity boundary condition was applied on the inflow/outflow boundaries in the airway. The controlled flow rates in right and left daughter airways are 55% and 45% of the flow rate in the trachea, respectively. The uniform velocity boundary conditions applied to the inflow/outflow boundary vary with time as a sinusoidal time function as shown in Figure 3.

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Re	1.75×10^3
Time steps /cycle	100
Time increment, Δt	0.05sec
Dimensionless time at each time increment, $t^* = 4 \nu \Delta t / D^2$	0.01

5. Results

The results are presented as the velocity profiles normalized by the maximum average local velocity in each branch. The results are taken from 8 cross-sections station throughout the model. The oscillatory flow simulation results in a

three-dimensional asymmetric bifurcation model at resting condition case are given in two separate parts, inspiration and expiration

5.1 Inspiration

An overview of respiratory flow features of the inspiration phase under the HFV condition is given in Figure 4. A uniform inlet velocity profile was imposed at the inflow boundary at the inlet parent airway. Flat velocity profiles with confined boundary layer thickness were found throughout the airway model and there was no secondary motion observed during the inspiration.

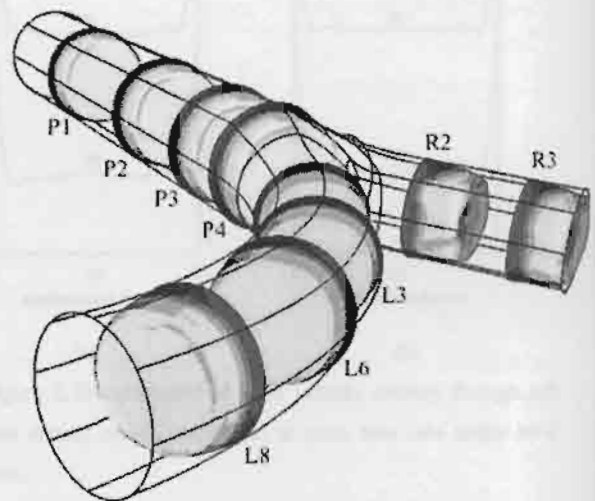


Figure 4 Inspiratory flow features at peak flow rate under HFV condition ($VT = 0.05L$, $f = 5\text{Hz}$, $Re = 4.37 \times 10^3$ and $\alpha = 11.87$)

The development of axial velocity profiles in the parent airway during inspiration was shown in Figure 4. The velocity profiles in the bifurcation plane (R-L) and normal plane to the bifurcation (A-P) are flat profiles with two-peak velocity profiles near the boundary layer throughout a straight pipe section of the parent airway. Note that, the boundary layer thickness in both planes is confined to be constant throughout the parent airway while the flow develops downstream.

When the flow enters the bifurcation, it experiences a change in flow direction and is split into two streams entering the right and left daughter branches. After the flow from the parent airway enters the right airway, the maximum axial velocity profiles in the bifurcation plane (R-L) are found near the outer wall of the bifurcation (not the inner wall as occurs under the resting condition and maximal exercise condition) as shown in Figure 5a. This can be explained that during the oscillatory cycle in HFV condition, the flow does not have sufficient time to become fully

developed. Therefore, the boundary layer thickness in this breathing condition is found approximately constant. In the normal plane to the bifurcation (A-P) as shown in Figure 6, axial velocity profiles are symmetric with deep flat profile at the centre of the flow and two-peak velocity profile near the boundary layer. These velocity profile patterns remain the same shape throughout the right daughter airway portion. Again, the boundary layer thickness found in right daughter airway in both planes does not change when the flow moves further downstream. Also, it was observed that the boundary layer thicknesses on both sides (inner wall and outer wall of the bifurcation) of the right daughter airway wall are equal.

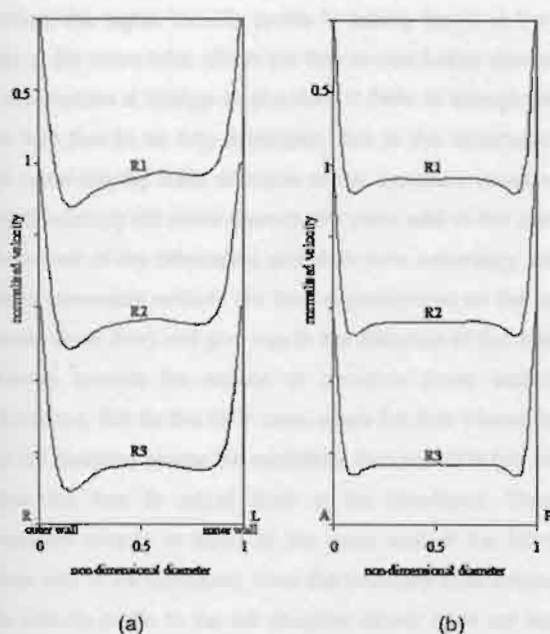


Figure 5 Development of axial velocity profiles through the right daughter airway during inspiration at peak flow rate under HFV condition.

Similar flow mechanisms occur in the left daughter airway. In the bifurcation plane (R-L), the axial velocity profiles are flat profiles with a maximum velocity profile near the outer wall of the bifurcation (inner wall of the curvature) as shown in Error! Reference source not found.a. When the flow develops downstream through the curved portion of the left daughter airway, similar flow patterns occur with peak velocity profiles next to the outer wall of the bifurcation. The boundary layer thickness is constant throughout the left daughter airway. In the normal plane to bifurcation (A-P), the velocity profiles are symmetric with peak velocity near the wall and a deep flat profile at the centre of the flow. The velocity profiles in this plane, again, retain the same flow features throughout the curved portion of the left daughter

airway with constant boundary layer thickness as shown in Figure 6.

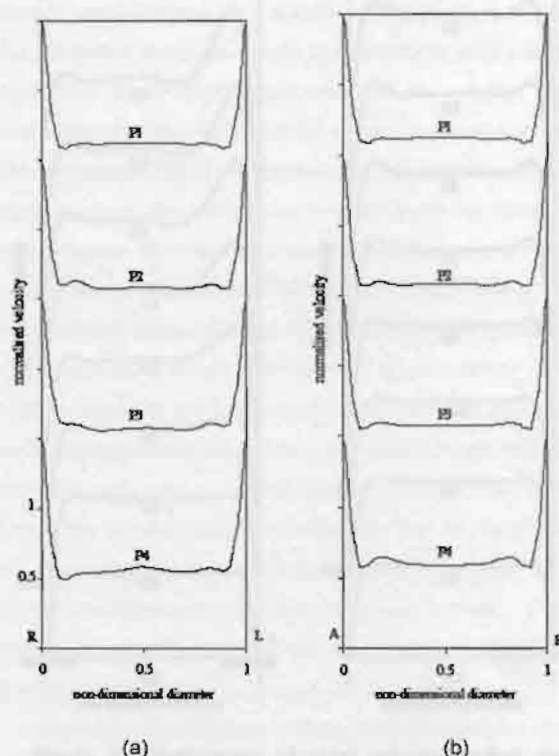


Figure 6 Development of axial velocity profiles through left daughter airway during inspiration at peak flow rate under HFV condition.

Note that, the velocity gradients of the axial velocity profiles throughout the airways during inspiration are confined to a narrow boundary layer and remain constant. It is found that secondary motion does not exist in the left daughter airway during the inspiration but very small secondary motion is observed at the inner wall of the bifurcation in the right daughter airway. However, the secondary velocity observed in the right daughter airway is extremely small compared to the axial velocity. Thus, the axial velocity profiles in the right daughter airway during inspiration are not distorted by the secondary motion. Therefore, the secondary motion effect observed at this side can be neglected.

5.2 Expiration

An overview of the expiratory flow of the HFV condition is given in Figure 7. The flow features shown here are taken from the peak expiratory flow rate during the respiratory cycle for the HFV condition. For expiration, uniform inlet velocity profiles were applied at the inflow boundaries in left and right daughter airways to correspond to flow rates of 45% and 55% of the trachea in left and right daughter airways.

When the flows in two daughter airways move downstream, the flow profiles remain flat with peak velocity profiles at the outer walls of the bifurcation. Figure 7 represents the development of the axial velocity profiles in left daughter airway. In the bifurcation plane (R-L), the axial flat profile at the inflow boundary in the left daughter airway develops further downstream along a curved portion of the left daughter airway with a prominent high peak velocity near the outer wall of the bifurcation (inner side of the curvature). This prominent velocity is more remarkable when the flow reaches the carinal ridge at station L1. The high peak velocity profiles found at the outer wall of the bifurcation in the curved portion of the left daughter airway is purely due to the curvature effect. When a uniform velocity profile enters a curved section, the higher velocity profile is initially found at the inner side of the curve tube. When the flow moves further downstream it experiences a change in direction. If there is enough time for the fluid flow to be fully developed, due to the centrifugal force the faster-moving fluids at inside of the curvature (outer wall of the bifurcation) will move towards the outer wall of the curvature (inner wall of the bifurcation) and then form secondary motions. These secondary motions are then superimposed on the primary motion (axial flow) and give rise in the distortion of the axial flow skewing towards the outside of curvature (inner wall of the bifurcation). But for the HFV case, when the flow moves through the left daughter airway the oscillatory time period is too small to allow the flow to adjust itself to be developed. Thus, the maximum velocity is found at the outer wall of the bifurcation (inner wall of the curvature). Also, the boundary layer thickness of the velocity profile in the left daughter airway does not increase throughout the daughter airway portion.

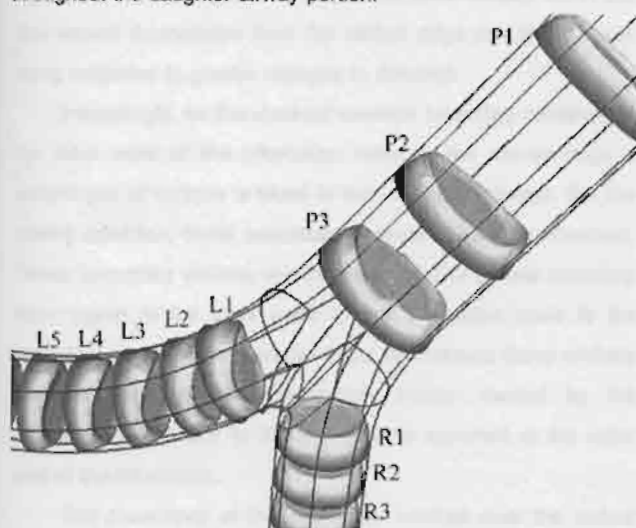


Figure 7 Expiratory flow features at peak flow rate under HFV condition ($VT = 0.05L$, $f = 5\text{Hz}$, $Re = 4.37 \times 10^3$ and $\alpha = 11.87$)

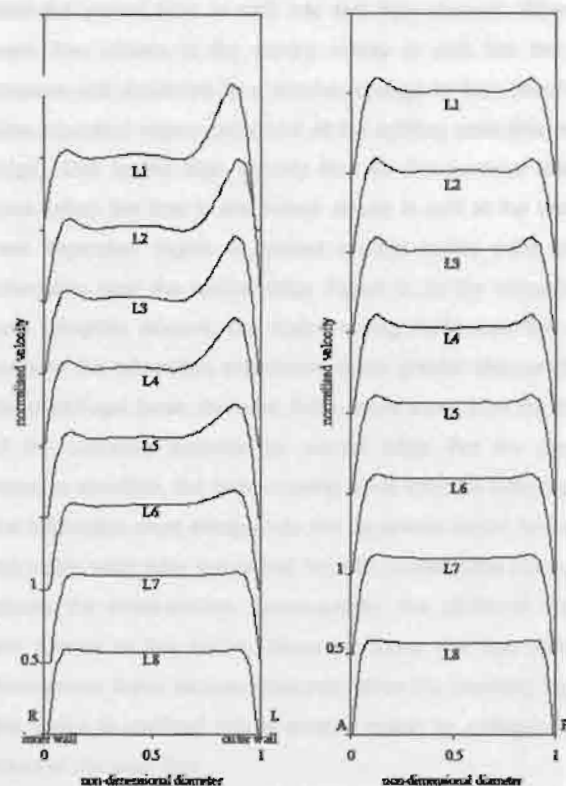


Figure 8 Development of axial velocity profiles through the parent airway during expiration at peak flow rate under HFV condition.

In the normal plane to the bifurcation (A-P), the axial velocity profiles begin with a flat profile. When the flow develops downstream a deep flat profile at the centre of the flow with a higher velocity profile near the wall is observed. Again, the boundary layer thickness does not increase when the flow moves along the curve section of the left daughter airway. The similar flow features are observed in right daughter airway. The flows from two daughter airways merge together at the bifurcation and then enter the parent airway at station P4. The asymmetric velocity profiles are found in the bifurcation plane (R-L) at station P4 (Figure 8). The velocity profile at this station has peak velocity near the wall. When the flow develops through station P3 the flow starts to be symmetrical in shape with higher velocity close to the wall and flat profile at the centre. When the flow reaches station P1 the velocity profile is completely symmetric with deep and flat profile at the centre and higher velocity peak near the wall. For the normal plane to the bifurcation, the flow profile at each station along the parent airway is symmetric. The flow starts with a flat profile at station P4 and then a profile with flat profile at the core of the flow and higher velocity near the wall at station P3, P2 and P1. The boundary layer thickness of the velocity profiles in the parent airway in both bifurcation plane and the plane normal to

the bifurcation does not increase and remain the same thickness. For the HFV condition, no secondary motion is observed in expiration throughout the airway.

6. Discussion

The respiratory flow simulation of the HFV condition at peak flow rate in this study was found to be sufficient agreed well with the experimental results given by [14]. The flow features obtained from the rest of the respiratory cycle during acceleration and deceleration are different from the flow features at the peak flow rate. During inspiration, the flow profiles at the equivalent flow rate between acceleration and deceleration within the same inspiration phase are significantly different.

It was found that during inspiration the distortion of axial velocity profiles for the resting condition and maximal exercise cases yielded similar flow patterns as shown in Figure 9.

In general, when the flow from the parent tube enters the daughter airways, the axial flows in both daughter airways for both cases are skewed towards the inner walls of the bifurcation (outer wall of the curvature). This conforms to steady flow in curved tube [16]. When the flow enters a curved section, the faster-moving fluid elements with maximum velocity profile are initially found at the inner wall of the curved (outer wall of the bifurcation). These faster-moving fluids laterally are subjected to a larger centrifugal force. These faster-moving fluids then move towards the outer wall of the bend (inner wall of the bifurcation) while the slower-moving fluids move towards the centre of curvature. This gives rise to superposition of the primary axial flow by the secondary motions. The secondary motion has more significant effect on the distortion of the axial velocity when the flow moves downstream from the carinal ridge due to the fluids being subjected to greater changes in direction.

Interestingly, for the maximal exercise breathing condition, at the inner walls of the bifurcation close to the carinal ridge a second pair of vortices is found in both daughter airways. For the resting condition, these secondary vortices cannot be observed. These secondary vortices are confined within a narrow boundary layer region at the inner walls of the bifurcation close to the carinal ridge. When the flow develops downstream these vortices disappear and the only secondary motion caused by the centrifugal forces due to the curvature is apparent at the outer wall of the bifurcation.

The occurrence of the secondary vortices near the carinal ridge for the maximal exercise condition is likely to be due to the escape of fluid particles from the main flow stream to a new boundary layer region formed at the carinal ridge when the flow

from the parent tube is split into two flow streams. When the main flow stream in the parent airway is split into two flow streams and subjected to a sudden change in flow direction, a new separated region can occur at the splitting point (the carinal ridge). Due to the high velocity flow for the maximal exercise case, when the flow in the parent airway is split at the carina a new separated region is formed on the inside walls of the bifurcation near the carinal ridge Figure 9. At the entrances of both daughter airways, the faster-moving fluids near the outer walls of the bifurcation experience much greater change due to the centrifugal force, then the fluids move away from the centre of the curvature towards the carinal ridge. For the maximal exercise condition, the faster-moving fluids from the outer walls of the bifurcation must escape into the separated region formed on the inside walls near the carinal ridge to conserve the momentum across the cross-section. Consequently, the additional vortices are formed at the carina. However, when the flow develops downstream these vortices disappear since the boundary layer at the carina is confined into a smaller region by stronger inertial effect of the axial flow.

In contrast, the respiratory flow patterns observed in the HFV condition case are totally different from those observed in the resting condition and maximal exercise condition. There is no secondary motion found in the parent and the left daughter airways during inspiration for the HFV condition. Only a less noticeable secondary motion is observed in the right airway. But this secondary motion has too small effect on the distortion of the axial velocity profiles. The axial velocity profiles in the bifurcation plane both in right and left daughter airways are rather flat with higher velocity near the outer walls of the bifurcation (not at the inner walls of the bifurcation as found in resting condition and maximal exercise condition). Note that, the boundary layer thickness for the HFV condition does not increase when the flow develops throughout the airway model. This suggests that the viscous effect plays less significant role in the respiratory flow under the HFV condition.

On expiration, due to the higher flow rate from the right daughter airway, the higher axial velocity profile in the bifurcation plane in the parent airway just downstream from the bifurcation is found near the right wall of the tube for the resting condition and the maximal exercise condition. The velocity profiles for the maximal exercise case are more flat than those found for the resting condition. This asymmetric flat profile persists until the flow reaches the outflow boundary in the parent airway.

Secondary motions are found in the parent airway during expiration for the resting condition case and maximal exercise

condition with double pairs of helical motions. The secondary motion has a more significant effect on the distortion of axial velocity profiles in the parent airway than the maximal exercise condition. This is indicated by the thicker boundary layer of the resting condition case compared to the maximal exercise condition.

For the HFV condition, the axial velocity profiles in the parent tube on expiration are significantly different from the profiles found in the resting condition and maximal exercise condition. The velocity profiles in the parent airway for the HFV condition are rather flat in the core of the tube with higher velocity near the both side of the tube. When the flow develops downstream, this higher peak velocity remains near the wall, but with less amplitude. Again, secondary motions are not found during expiration for the HFV condition. Also, the boundary layer thickness throughout the system is very thin and consistent. It can be said that the viscous effect does not play a significant role on respiratory flow patterns in the HFV condition, even at the low Reynolds number where the viscous effect should be taken into account in considering the flow behavior.

Comparing to the resting condition and maximal exercise case, where the breathing frequencies for both cases are small enough to allow the flow to adjust itself to develop, it was found that the flow profiles are strongly dependent on the flow rate (Reynolds number) rather than the breathing frequency (Womersley number). With the increasing Reynolds number the boundary layer thickness for the resting condition decreases. The boundary layer thickness for the maximal exercise condition is influenced by the Reynolds number in the same manner. The boundary layer thickness for the resting condition is thicker than in the maximal exercise condition. This indicates that the inertia force becomes even greater than the viscous force for the maximal exercise case as compared to the resting condition.

It can be concluded that Reynolds number is the important parameter influencing the respiratory flow as long as the breathing frequency is small enough to allow the flow to adjust itself to develop. This also suggests that the flow in the low breathing frequency range is affected by viscous effect, thus the geometry of the airway is of important. In other words, the respiratory flow is dependent on the Dean number when the Womersley number is small (small breathing frequency). However, the viscous effect becomes less important when the breathing frequency becomes high, such as high-frequency ventilation. Thus the geometry effect can be neglected for the HFV condition and only the frequency parameter (Womersley number) plays a significant role on the respiratory flow patterns.

7. Conclusions

It was found that the numerical results of the HFV respiratory flow in an asymmetric single bifurcation of the central airway at peak flow rate are sufficient to give a number of results which both qualitatively and quantitatively agree well with the experimental results given by [14].

The results also showed that the respiratory flow under HFV condition are dependent only on the unsteadiness effect, while the respiratory flows at resting condition and maximal exercise condition are dominated by the convective and viscous effects. Hence, the realistic geometry of the airway is important to get the realistic results of the airflow patterns for the resting condition and exercise condition. While for the HFV condition, the geometry effect becomes less important. It also was found that the respiratory flow under the high-frequency ventilation behaves in the similar way as in the straight tube.

The secondary flow motions were observed in both daughter airways during inspiration and in the parent airway during expiration phase under both resting and maximal exercise conditions. Whereas the secondary flow under the HFV condition was not obtained. The numerical flow information in the study is useful to track on the particle deposition during breathing cycle in order to investigate the efficacy of the pharmaceutical which most are in the form of aerosol particle.

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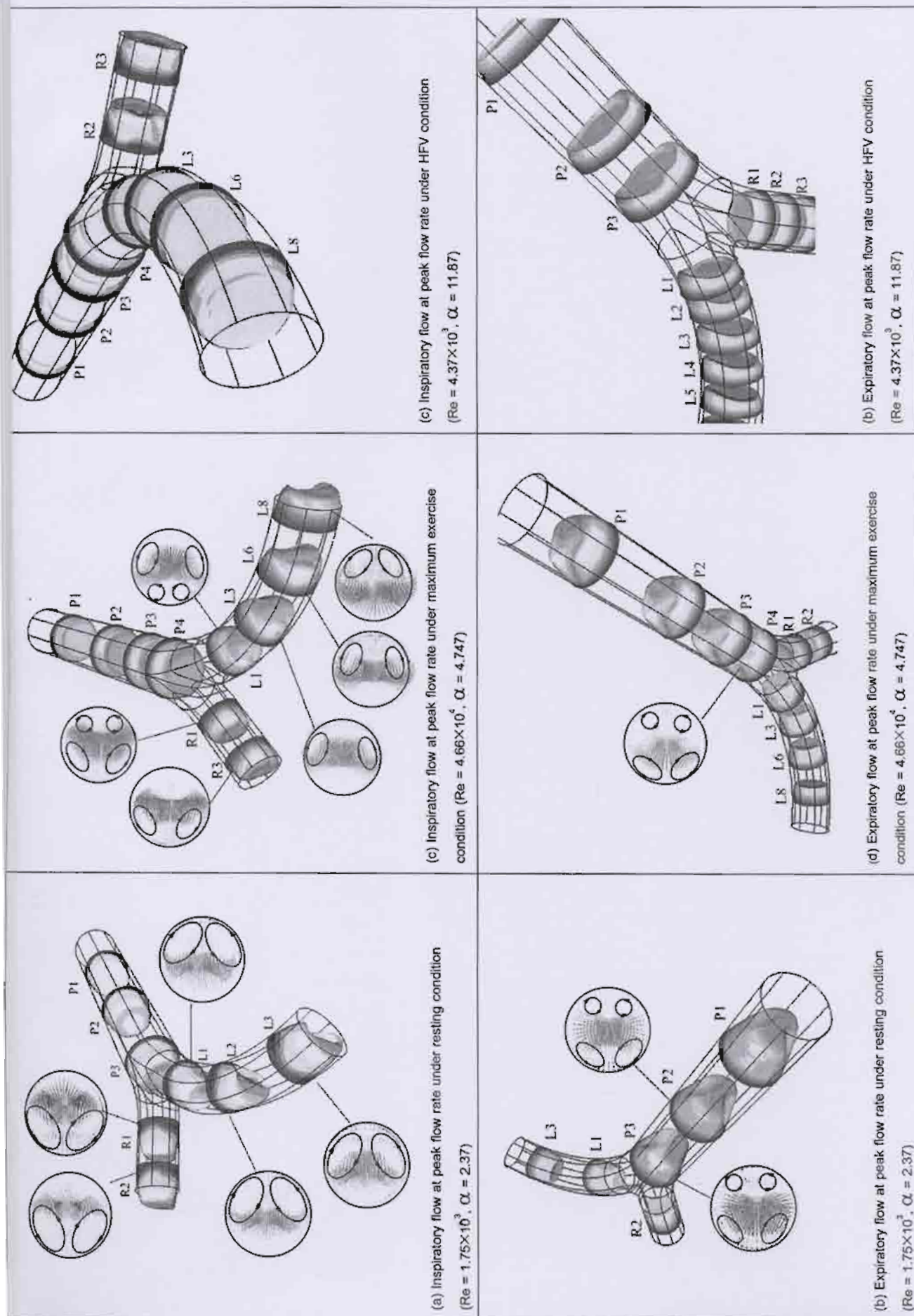


Figure 9 Respiratory flow features at peak flow rate under different breathing conditions